TECHNICAL NOTE

Optimal Imaging Parameters for Readout-segmented EPI of the Temporal Bone

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Readout-segmented echo planar imaging (rs-EPI) is a form of multi-shot EPI. rs-EPI is affected less by susceptibility artifacts than single-shot EPI (ss-EPI), which is widely used for diffusion-weighted imaging, so rs-EPI is expected to produce less image distortion. In this study, we compared rs-EPI and conventional ss-EPI of the temporal bone region, which contains abundant amounts of air and frequently exhibits changes in magnetic susceptibility. In addition, we used a phantom to determine the optimum rs-EPI acquisition conditions for clinical use and investigated the clinical utility of rs-EPI in 20 patients (8 men, 12 women, mean age, 54.3 ± 16.7 -years-old) with cholesteatoma (mean apparent diffusion coefficient on ss-EPI, $0.88 \times 10^{-3} \pm 0.18 \text{ mm}^2/\text{s}$). The images of the temporal bone region produced using rs-EPI exhibited less distortion than those obtained with ss-EPI ($P < 0.05$).

Keywords: *distortion, readout-segmented EPI (rs-EPI), single-shot EPI (ss-EPI), susceptibility, temporal bone region*

Introduction

Diffusion-weighted imaging (DWI) is widely used to examine all regions of the body and is an essential method for imaging the head region.¹ The structures found in the head are complex. The temporal bone region contains a lot of air, and susceptibility artifacts, such as image distortion and chemical shifts, have been reported during single-shot EPI (ss-EPI) of this region. Although these problems have recently been improved using ss-EPI in combination with parallel imaging, the improvement has been limited.^{2–4} Multi-shot EPI (ms-EPI), in which the k-space is divided in the phase-encoding direction, is considered to shorten the duration of echo reading. However, bulk motion between shots may produce phase changes, which often produce artifacts.^{5,6} To counter these issues, Porter and colleagues designed a method for segmenting the k-space in the readout direction and reading the sig-

nals in each segment⁷ (Fig. 1). Data are sampled at fewer points in the readout direction in this method, so the echo space (ES) is shortened, which reduces the effect of susceptibility artifacts on image quality.

We determined the optimum conditions for acquisition of readout-segmented echo planar imaging (rs-EPI) for clinical imaging of the temporal bone region and investigated the clinical utility of the technique using a 3-tesla 32-channel coil.

Materials and Methods

rs-EPI employs the 2-spin echo-type data collection method, in which both imaging and navigator echo data are collected. In this method, a 180° refocusing pulse is reapplied after the acquisition of imaging echo data to collect the navigator echo data and counteract the influence of subject movement (Fig. 2).

Phantom study

All studies were performed using a 3T magnetic

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Fig. 1. k-space filling method on 5-shot readout-segmented echo planar imaging (rs-EPI).

Fig. 2. Readout-segmented echo planar imaging (rs-EPI) pulse sequence chart.

resonance (MR) unit (MAGNETOM Verio, Siemens, Erlangen Germany) with a 32-channel phased-array coil. Sensitivity correction was performed using a pre-scan normalization filter. The ss-EPI acquisition sequence was: repetition time (TR), 8800 ms; echo time (TE), 84 ms; echo space, 1.04 ms; b-factor, 800 s/mm²; matrix size, 130 × 130 (zero interpolation [ZIP] to 260); bandwidth, ±1068 Hz; number of slices, 30; slice thickness, 1.5 mm, without gaps (voxel size, 1.7 × 1.7 × 1.5 mm); width of square field of view (FOV), 220 mm; acceleration factor (AF), 4; acquisition time, 3 min 49 s; and number of excitations (NEX), 5. In addition, phase encoding was performed in the anterior-to-posterior (A-P) direction, generalized autocalibrating partially parallel acquisition (GRAPPA, a parallel imaging technique) was employed, and motion-probing gradient (MPG) pulses were applied to all 3 axes.

During the rs-EPI, the matrix size, AF, bandwidth, and number of segments were varied to determine the optimal acquisition conditions. Specifically, the rs-EPI acquisition conditions were: TR, 6500 ms; TE, 5.59 to 70 ms; echo space, 0.32 ms; b-factor, 800 s/mm²; matrix size, 192 × 192 or 226 × 226; bandwidth, ±359 to 723 Hz; number of slices, 30 slices; slice thickness, 1.5 mm, without gaps (voxel size, 1.1 × 1.1 × 1.5 mm and 1.0 × 1.0 × 1.5 mm); width of the square FOV, 220 mm; AF, 2 or 4; number of k-space segments, 9, 11, or 19; acquisition time: 5 min 1 s to 10 min 13 s; and NEX, one. Phase encoding was performed in the A-P direction, GRAPPA was employed, and MPG pulses were applied to all 3 axes. The rs-EPI scans were performed in a temperature-controlled room at 21°C.

We employed a custom-made phantom for these experiments, adding to the phantom's sample holder either acetic acid (T_1 value, 797 ms; T_2 value, 234 ms; apparent diffusion coefficient (ADC), 1.24×10^{-3} mm²/s) because it produces a similar signal to cholesteatoma on ss-EPI (mean ADC value, 0.88×10^{-3} mm²/s ± 0.18; n = 20) or distilled water (T_1 value, 3782 ms; T_2 value, 1947 ms; ADC, 1.87×10^{-3} mm²/s) because it was assumed that a lot of fluid would be present in the temporal bone region. The sample holder had an inner diameter of 10 mmφ and was inserted into a special type of phantom containing polyvinyl alcohol gel (150 × 150 mmφ, Nikko Fines Industries, Tokyo, Japan). We compared the signal-to-noise ratios (SNR) produced during imaging of the phantom between ss-EPI and rs-EPI. Images were acquired twice, and we calculated the SNR according to the difference method using: $SNR = \text{Signal} / (SD_{\text{sub}} / \sqrt{2})$. "Signal" represents the mean signal intensity value derived from 2 images acquired under the same conditions in a region of interest (ROI) measuring 7 × 7 pixels,⁸ and SD_{sub} represents the standard deviation of the latter value.

To measure the distortion ratio (DR), we placed acetic acid in a straw with an inner diameter of 4, 7, or 10 mmφ in the phantom's sample holder. The diameter of the straw on T_2 -weighted imaging performed in the phase direction (TR, 4000 ms; TE, 92 ms) was designated as B, and that observed on ss-EPI after the application of MPG pulses to all 3 axes or on the isotropic DWI obtained with rs-EPI was designated as A. Then, the distortion ratio (DR) was calculated using the equation: $DR = A/B$.

To measure the resolution of the images obtained with the 2 techniques, we placed 2 blocks of pins, each containing 4 lines of 5 pins, in the phantom at a right angle to each other. Diameters of all the pins in each line were the same, and the 4 lines contained pins with inner diameters of 2.0, 1.0, 0.75, and 0.5

Fig. 3. Comparison of single-shot (ss-) and readout-segmented echo planar imaging (rs-EPI) (shots, fixed to 11) images of resolution phantom with changes in the matrix size and acceleration factor. (A) ss-EPI. (B) rs-EPI at matrix size, 192×192 and acceleration factor, 2. (C) rs-EPI at matrix size, 192×192 and acceleration factor, 4. (D) rs-EPI at matrix size, 226×226 and acceleration factor, 2. (E) rs-EPI at matrix size, 226×226 and acceleration factor, 4.

mm. MPG pulses were applied to the 3 axes during the ss-EPI and the isotropic DWI obtained with rs-EPI (Fig. 3). When pins were correctly detected on the images acquired in the phase direction, the resolution of those with inner diameter of 2.0 mm was one, of 1.0 mm, two, of 0.75 mm, three, and of 0.5 mm, four. Seven radiologists, each with more than a year's experience in MR imaging, performed the visual evaluations. The significance of differences was analyzed using the Mann-Whitney U-test (Fig. 3D).

Clinical study

The clinical ethics committee of our hospital approved the study, and subjects consented to participation. Subjects were 20 patients (8 men, 12 women, mean age, 54.3 ± 16.7 years) with cholesteatoma (14 on the right, 8 on the left). All subjects

underwent imaging between November 2012 and May 2013.

In this clinical study, the ss-EPI images were acquired using the following sequence: TR, 8800 ms; TE, 84 ms; echo space, 1.04 ms; b-factor, 800 s/mm²; matrix size, 130×130 (zero interpolation [ZIP] to 260); bandwidth, ± 1068 Hz; number of slices, 30; slice thickness, 1.5 mm without gaps (voxel size: $1.7 \times 1.7 \times 1.5$ mm); width of square FOV, 220 mm; AF, 4; acquisition time, 3 min 49 s; and NEX, 5. In addition, phase encoding was performed in the A-P direction, GRAPPA was employed, and MPG pulses were applied to all 3 axes.

The rs-EPI acquisition sequence was: TR, 6500 ms; TE, 8.70 ms; echo spacing, 0.32 ms; b-factor, 800 s/mm²; matrix size, 226×226 ; bandwidth, ± 572 Hz; number of slices, 30; slice thickness, 1.5 mm, without gaps (voxel size: $1.0 \times 1.0 \times 1.5$ mm); width of square FOV, 220 mm; AF, 2; number of k-space segments, 11; acquisition time, 6 min 6 s; and NEX, one. In addition, phase encoding was performed in the A-P direction, GRAPPA was employed, and MPG pulses were applied to all 3 axes.

For each of the acquired ss-EPI images and the isotropic DWI images obtained with rs-EPI, we calculated the contrast-to-noise ratio (CNR) of the lesion relative to the brainstem using the equation: $CNR = (S1 - S2) / \{(SD1^2 + SD2^2) / 2\}^{1/2}$, in which S1 represents the mean signal intensity value for ROI1 and S2, that for ROI2, and SD1 represents the standard deviation of the mean signal intensity value for ROI1 and SD2, that for ROI2. ROI1 and ROI2 both measured 50 pixels.⁹

Two neuroradiologists with more than 20 years' experience used a 5-point system to grade the ability of rs-EPI to visualize lesions (relative to that of ss-EPI), scoring images that exhibited marked improvement as excellent, 5; slight improvement, good, 4; similar quality, equal, 3; local failure to visualize the lesion, fair, 2; and complete failure to visualize the lesion as poor, one. The significance of differences was analyzed using the Mann-Whitney U-test.

Results

Phantom study

Figure 4 shows the SNR values exhibited by the acetic acid and distilled water in the phantom's sample holder (inner diameter, 10 mm ϕ) on ss- and rs-EPI. Both substances exhibited higher SNR on ss-EPI than on rs-EPI at matrix sizes of 226×226 (Fig. 4A) and 192×192 (Fig. 4B). On rs-EPI, the SNR rose as the number of segments increased and decreased as the AF increased. No sig-

Fig. 4. Signal-to-noise ratio (SNR) of acetic acid versus distilled water in each sequence. (A) Matrix size, 226×226 . (B) Matrix size, 192×192 .

nificant difference in the SNR was observed between the images obtained using matrix sizes of 192×192 and 226×226 .

In the next experiment, we placed straws with inner diameters of 4, 7, and 10 mm ϕ that contained acetic acid into the sample holder of the phantom and measured the distortion ratio (DR). Figure 4 shows the results. Compared with the DR observed on ss-EPI, which were taken as 1.0 (no distortion), the DR observed on rs-EPI became closer to 1.0 as the diameter of the straws increased at matrix sizes of both 226×226 (Fig. 5A) and 192×192 (Fig. 5B). The change ($1.0 - 0.86 = 0.14$) was particularly large for the 4-mm straw. The DR was lower on each rs-EPI sequence at all diameters (4, 7, and 10 mm ϕ) and near 1.0 in all of these sequences. In a comparison between matrix sizes of 226×226 and 192×192 , all the rs-EPI sequences produced similar DRs. Figure 3 shows images of the phantom used for the resolution analysis (resolution phantom). Figure 6 shows the results of the visual evaluations of resolution performed by the 7 radiologists using the resolution phantom. The mean resolution score was low (0.29) for the ss-EPI images and

Fig. 5. Distortion ratio (DR) values of acetic acid-containing straws with inner diameters of 4, 7, and 10 mm ϕ . (A) Matrix size, 226×226 . (B) Matrix size, 192×192 .

high for each rs-EPI sequence (1.0 or higher), with significant differences observed between the ss-EPI sequence and each rs-EPI sequence ($P < 0.05$). In a comparison among the rs-EPI sequences, the sequence involving a matrix size of 226×226 and an AF of 2 exhibited the highest mean value (1.71). When the AF was increased to 4, the mean resolution score decreased, though not significantly, to 1.57 (matrix size, 226×226). The mean resolution score also decreased when the matrix size was decreased to 192×192 . The mean resolution score decreased further when the AF was increased to 4, as was observed at a matrix size of 226×226 . A significant difference was only detected between the sequence involving a matrix size of 226×226 and an AF of 2 and the sequence involving a matrix size of 192×192 and an AF of 4 ($P < 0.05$).

Clinical study

Figure 7A shows the CNR of 20 cholesteatoma lesions (relative to the brainstem) on ss-EPI and isotropic DWI acquired using rs-EPI. CNRs of the le-

Fig. 6. Visual evaluation by 7 radiologists of single-shot (ss)- and readout-segmented echo planar imaging (rs-EPI) (shots, fixed to 11) images of resolution phantom with changes in the matrix size and acceleration factor. n.s., not significant; * $P < 0.05$; ** $P < 0.005$.

sions were significantly higher on rs-EPI than ss-EPI ($P < 0.05$). The visual evaluation scores were also significantly better for the rs-EPI images than the ss-EPI images ($P < 0.05$) (Fig. 7B).

Following are typical clinical cases. Patient 1 was a 58-year-old man who developed a recurrent right cholesteatoma following surgery. The lesion could not be clearly identified on ss-EPI but was successfully visualized and exhibited high intensity and reduced distortion on rs-EPI (Fig. 8). Patient 2 was a 65-year-old woman with a right cholesteatoma. The lesion could be identified on ss-EPI, but the lesion was less distorted and its morphology was easier to visualize on rs-EPI. Furthermore, the rs-EPI image exhibited high resolution (Fig. 9).

Discussion

We performed a basic investigation and assessment of the clinical utility of rs-EPI. When sample holders (inner diameter, 10-mm ϕ) containing acetic acid or distilled water were imaged using rs-EPI and ss-EPI, the highest SNR observed on rs-EPI (19 segments; acquisition time 10 min 13 s) was lower than that on ss-EPI. Because the echo space remained constant, the SNR was improved by increasing the number of segments. The latter was achieved by prolonging the readout time and reducing the bandwidth. However, this prolonged the acquisition time, which is not appropriate for clinical

examinations. On rs-EPI, the SNR did not increase when the matrix size was reduced to 192 $\times$ 192, which might have been due to an increase in the bandwidth. The SNR did not decrease when the matrix size was set to a high resolution of 226 $\times$ 226. Thus, the latter matrix size was considered useful to improve images by leading to an increased number of sampling points in both readout directions.

In the phantom and clinical studies, SNR and CNR were measured using a 32-channel phased-array coil. The use of a prescan normalize filter for sensitivity correction was considered acceptable because the same filter was applied to both signals and noise. As for the issues of heterogeneous sensitivity and parallel imaging, we considered our phantom measurements accurate because we employed the subtraction mapping method described by Imai's group⁸ and adopted an ROI size of 7 $\times$ 7 pixels.

Heterogeneous sensitivity and parallel imaging were also present during the CNR measurements performed in the clinical study, but these problems were taken into consideration by the formula used to calculate the CNR of the clinical images: $CNR = (S1-S2)/\{(SD1^2+SD2^2)/2\}^{1/2}$. According to the study reported by Ogura and associates,⁹ stable values with a normal distribution can be obtained using an ROI size of 50 pixels, the same as that we used in the present study.

We also compared the DR of acetic acid-containing straws with inner diameters of 4, 7, or 10 mm ϕ

Fig. 7. Evaluation of 20 patients with cholesteatoma on single-shot (ss)- and readout-segmented echo planar imaging (rs-EPI). **(A)** Comparison of contrast-to-noise ratio (CNR) of cholesteatoma versus brainstem. CNR significantly increased on rs-EPI. $*P < 0.05$. **(B)** Visual evaluation by 2 neuroradiologists. The score significantly increased on rs-EPI. $*P < 0.05$.

between ss-EPI and rs-EPI. On rs-EPI, the DR was close to 1.0 (representing the absence of distortion), even at an inner diameter of 4 mm. On ss-EPI, the DR changed depending on the inner diameter of the straw, which might have been due to differences in magnetic susceptibility. Because the magnetic dipole is proportional to the magnetization of the 2 poles and the distance between them, its influence might have been large, even at identical levels of magnetic susceptibility. On rs-EPI, when the matrix size was increased from 192×192 to 226×226 , i.e., when the resolution was improved, the DR remained near 1.0. This suggests that increasing the resolution prolongs the readout time. Although this would have induced image distortion on ss-EPI, the absolute amount of distortion would have been decreased on rs-EPI by the reduction in pixel size, even though the readout time was extended.¹⁰ Therefore, high resolution images can be acquired by setting the matrix size to 226×226 . During the measurements obtained with the resolution phantom, the highest resolution score was obtained at a matrix size of 226×226 and an AF of 2, which agrees with the findings of the DR experiment. The GRAPPA parallel imaging technique, which has been suggested to shorten the readout time and reduce distortion, is applicable to rs-EPI. However, when the AF was set to 4, the SNR decreased markedly and visual resolution also fell.

Based on the above findings, the use of GRAPPA with an AF of 2, 11 segments, and an acquisition time of 6 min 6 s might be appropriate because susceptibility artifacts and distortion are less severe in these conditions.

In the clinical cases, lesional CNR (relative to the

Fig. 8. Right cholesteatoma in a 58-year-old man. **(A)** Single-shot echo planar imaging (ss-EPI). **(B)** Readout-segmented (rs)-EPI. Cholesteatoma was more clearly visualized on rs-EPI (arrow).

Fig. 9. Right cholesteatoma in a 65-year-old woman. (A) Single-shot echo planar imaging (ss-EPI). (B) Readout-segmented (rs)-EPI. Visualization of cholesteatoma was improved on rs-EPI, and signal variation in the cholesteatoma was also visualized (arrow).

brainstem) and visual evaluation scores were better on rs-EPI than ss-EPI ($P < 0.05$). This suggested that although the SNRs of the rs-EPI images were decreased, they exhibited less severe susceptibility artifacts and distortion, resulting in improved visualization of the target lesions.

Morelli,¹¹ Holdsworth,¹² and Naganawa and associates¹³ reported the acquisition of high resolution diffusion images and improved visualization of the brainstem and cranial nerves during studies of the head in patients with stroke. Whereas Naganawa's group set the voxel size on rs-EPI at $0.5 \times 0.6 \times 4$ mm, we set it at $1.0 \times 1.0 \times 1.5$ mm, which reduced the slice thickness, during imaging using a matrix size of 226×226 because the target lesions were cholesteatomas. As a result, we observed improved visualization of cholesteatoma lesions.

However, the increase in motion artifacts, decrease in SNR, and inability to change the TE or bandwidth of the rs-EPI sequence freely due to prolongation of the acquisition time are problematic. Therefore, further improvement of the rs-EPI sequence is required.

Conclusion

We segmented the k-space in the readout direction and performed a basic analysis of diffusion imaging using a new readout-segmented EPI (rs-EPI) protocol in which the signal for each segment was read. Then, we investigated the utility of this method in the clinical setting. In our phantom experiments, the diffusion images acquired after ap-

plying MPG pulses to the 3 axes exhibited less severe susceptibility artifacts and distortion. The acquisition conditions were optimized based on the results of the latter experiments. During clinical imaging of 20 patients using our new protocol, the CNR of the target cholesteatoma lesions (relative to the brainstem) were increased. Thus, our new protocol demonstrated a markedly improved ability to visualize such lesions. The sequence described in this report might replace previous methods.

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