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Half-Fourier Single-Shot STEAM MRI

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As a high-speed imaging technique based on stimulated echoes single-shot STEAM MRI is insensitive to chemical shift artifacts and magnetic susceptibility differences. The achievable signal-to-noise ratio (SNR) is limited by the fact that high flip angles of the read-out excitation pulses cause a steep decay of the stimulated echo train and therefore degrade the point-spread function of the resulting image. The present work investigates the use of half-Fourier phase encoding which weakens the flip angle constraint by reducing the number of necessary excitations. Single-shot STEAM MRI of the normal human brain at 2.0 T demonstrates that half-Fourier versions either reduce the measurement time by almost a factor of two without sacrificing the SNR or increase the SNR by about 40% while keeping the measurement time constant. Magn Reson Med 47:611–615, 2002. © 2002 Wiley-Liss, Inc.

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Single-shot STEAM (stimulated echo acquisition mode) MRI (1) is a high-speed imaging technique with subsecond acquisition times based on RF refocused echoes. The resulting images are insensitive to resonance offset effects from chemical shifts or magnetic susceptibility differences and offer robust alternatives to echo-planar imaging (EPI), for example, for diffusion-weighted MRI (2). A drawback of single-shot STEAM MRI in comparison with EPI is a slightly longer measurement time and a reduced signal-to-noise ratio (SNR). The latter is not only because stimulated echoes refocus only half of the magnetization, but mainly because the flip angle of the read-out RF pulses is limited by its degradative effect onto the point-spread function (PSF) via the attenuation of the echo train (2). As the signal decrease in a stimulated echo sequence is dominated by T_1 relaxation, some SNR can be regained by using a reduced bandwidth, which lengthens the overall acquisition period without significantly enhancing relaxation losses.

Half-Fourier phase-encoding (3) omits the acquisition of one-half of the *k*-space except for a few central lines: following 2D Fourier transformation of the complex raw data, a phase correction based on the central *k*-space lines shifts the desired image information to the real data part, while the imaginary part is discarded. Half-Fourier phase encoding can therefore be exploited to reduce the acquisition time or improve the performance of single-shot imaging techniques. For example, it has already been used in conjunction with RARE (single-shot fast spin-echo) sequences to lower the inherent RF power deposition (4) and EPI to reduce the degree of (differential) T_2^*-weighting (5).

In most applications half-Fourier phase encoding is at the expense of SNR due to the lower number of acquired signals. However, specific sequences may still benefit when taking relaxation effects into account. For instance, half-Fourier Burst imaging leads to a better SNR than full-Fourier Burst because it minimizes T_2-related signal attenuation prior to the acquisition of the intensity-relevant central *k*-space lines (6). Here, a similar improvement is aimed at by combining half-Fourier phase encoding with single-shot STEAM MRI. For a given quality of the PSF, the reduced number of required Fourier lines or read-out excitations allows the use of larger flip angles, which compensate for the expected SNR loss. In fact, the SNR may even be improved when investing the reduced imaging time into a narrower bandwidth, i.e., longer data acquisition intervals.

THEORY

Figure 1 shows the basic RF and gradient pulse sequence for single-shot STEAM MRI. Following the preparation of slice-selective longitudinal magnetization by the leading two RF pulses, the repetitive read-out intervals involve low flip angle RF excitations to generate differently phase encoded stimulated echoes for single-shot coverage of the *k*-space. During this TR interval the available longitudinal magnetization is reduced by $\cos\alpha$ due to RF excitation and by $\exp(-\mathrm{TR}/T_1)$ due to T_1 relaxation. Consequently, the amplitudes of successive stimulated echoes decay by $\cos\alpha \cdot \exp(-\mathrm{TR}/T_1)$ during sequence progression.

Full-Fourier Phase Encoding

To maximize the SNR of single-shot STEAM MRI it is advantageous to use a centric reordered phase encoding scheme because it first acquires the low spatial frequencies that encode for most of the image intensity. Starting in the center, the scheme alternates Fourier lines in each half of the *k*-space, so that the amplitudes of two neighboring lines in *k*-space differ by a factor of $[\cos\alpha \cdot \exp(-\mathrm{TR}/T_1)]^2$ as shown in Fig. 2a (dashed lines for two different flip angles). A higher flip angle α gives rise to a larger echo amplitude of the central *k*-space lines but also leads to a steeper signal decay along the echo train, i.e., the phase encoding dimension of *k*-space. This effect broadens the resulting PSF of the image, which therefore becomes blurred along the phase encoding dimension. The full width at half-maximum of the PSF in the phase encoding direction ($\mathrm{FWHM}_{\mathrm{pe}}$) depends on the distance of neighboring lines in *k*-space, i.e., the field-of-view ($\mathrm{FOV}_{\mathrm{pe}}$), and is given by (2):

$$\mathrm{FWHM}_{\mathrm{pe}} = \frac{2}{\pi} \cdot \mathrm{FOV}_{\mathrm{pe}} \cdot (\mathrm{TR}/T_1 - \ln\cos\alpha). \qquad [1]$$

Thus, for a desired $\mathrm{FWHM}_{\mathrm{pe}}$ or pixel resolution the flip angle for full-Fourier phase encoding α_{FF} is limited to:

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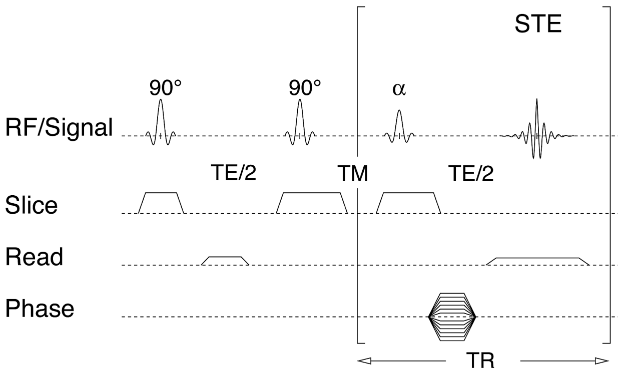

FIG. 1. Basic RF and gradient pulse sequence for single-shot STEAM MRI. The used implementation involved crusher gradients, RF spoiling for the low flip angle excitations, and a phase-encoding rewinder gradient in the TR interval.

$$\alpha_{FF} = \arccos\left(\ln\left(TR/T_1 - \frac{\pi}{2}\frac{FWHM_{pe}}{FOV_{pe}}\right)\right). \quad [2]$$

For example, for covering a 160 mm FOV with 80 stimulated echoes at a $FWHM_{pe}$ of 2 mm, the flip angle should not exceed 9.4° assuming a repetition time of TR = 6 ms (image pixel bandwidth BW = 280 Hz) and a spin-lattice relaxation time of T_1 = 1000 ms (corresponding to gray matter at 2.0 T).

Half-Fourier Phase Encoding

Half-Fourier phase encoding skips half of k-space except for a few central lines that are required for phase correction. It therefore considerably reduces the acquisition time, but in most circumstances also sacrifices SNR because of the lower number of acquired echoes N with SNR $\propto \sqrt{N}$. On the other hand, half-Fourier single-shot STEAM MRI benefits from an ascending phase encoding scheme starting in the center of k-space, which ensures that neighboring k-space lines directly follow each other in time, i.e., are acquired in direct succession. Therefore, their echo amplitudes only differ by a factor of $\cos\alpha \cdot \exp(-TR/T_1)$, which reduces the resulting $FWHM_{pe}$ of the PSF by a factor of two:

$$FWHM_{pe} = \frac{1}{\pi} \cdot FOV_{pe} \cdot (TR/T_1 - \ln\cos\alpha) \quad [3]$$

with the corresponding limiting flip angle α_{HF} given by:

$$\alpha_{HF} = \arccos\left(ln\left(TR/T_1 - \pi\frac{FWHM_{pe}}{FOV_{pe}}\right)\right). \quad [4]$$

As a consequence, for a given repetition interval TR, half-Fourier phase encoding allows for a larger flip angle than full-Fourier phase encoding according to:

$$\cos\alpha_{HF} \cdot \exp(-TR/T_1) = \cos^2\alpha_{FF} \cdot \exp(-2TR/T_1) \quad [5]$$

or:

$$\cos\alpha_{HF} = \cos^2\alpha_{FF} \cdot \exp(-TR/T_1). \quad [6]$$

The corresponding signal gain can even overcompensate the $\sqrt{N}$ loss, so that half-Fourier single-shot STEAM MRI may benefit from an improved SNR relative to full-Fourier phase encoding. For the above-mentioned example, the flip angle can be increased from 9.4° to 14.7°, yielding an SNR gain of more than 10% with a simultaneous reduction of the acquisition time from 492 ms to 276 ms.

Alternatively, it is also possible to prolong the TR interval and retain the acquisition time needed for full-Fourier phase encoding. The additional time per TR may be invested in a reduced image pixel bandwidth, i.e., a longer acquisition period per stimulated echo, aiming at a further increase of the SNR. For the given example, an image pixel bandwidth of BW = 120 Hz instead of 280 Hz together

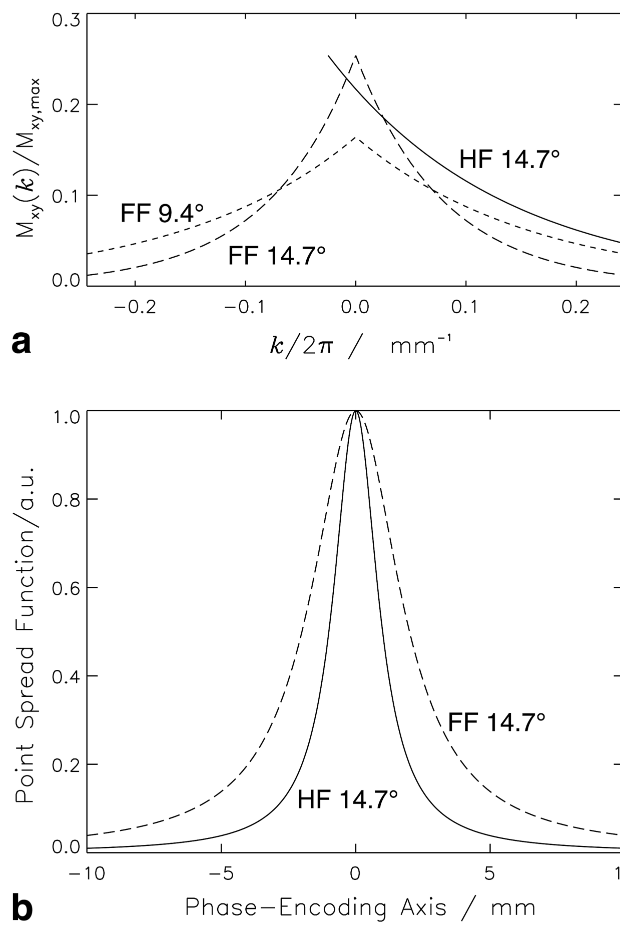

FIG. 2. **a:** Signal decay of successive stimulated echoes, i.e., Fourier lines along the phase encoding dimension of k-space, in single-shot STEAM MRI for full-Fourier (FF) phase encoding with a flip angle of 9.4° and 14.7° (dashed lines) as well as for half-Fourier (HF) phase encoding with a flip angle of 14.7° (solid line). **b:** Point-spread functions (PSF) for FF (dashed line) and HF (solid line) phase encoding with a flip angle of 14.7°. The HF curve coincides with the PSF for FF phase encoding and a flip angle of 9.4°.

with a flip angle of 13.6° is estimated to result in a 50% gain of the SNR without prolonging the acquisition time.

Signal-to-Noise Ratio

As discussed in the previous section, half-Fourier single-shot STEAM MRI does not automatically lead to a penalty in SNR by the reduced number of acquired signals. However, the achievable compensation by an increased flip angle α_{HF} depends on the $FWHM_{pe}$ of the PSF and the FOV_{pe} (both via α_{FF}) as well as on TR and T_1.

In order to estimate the range of tissue and sequence parameters for which single-shot STEAM sequence are likely to benefit from half-Fourier phase encoding, two simplifications are used: 1) the Fourier lines for phase correction, which are acquired prior to the central k-space line, are neglected (i.e., $N_{HF} = N_{FF}/2$), and 2) the image intensity I is assumed to be dominated by the signal amplitude of the central k-space line (i.e., $I \propto \sin \alpha$). Thus, half-Fourier phase encoding is expected to offer a better SNR than full-Fourier phase encoding if:

$$\frac{1}{\sqrt{2}} \cdot \sin \alpha_{HF} > \sin \alpha_{FF}. \qquad [7]$$

Equation [7] can be transformed to:

$$\frac{FWHM_{pe}}{FOV_{pe}} < \frac{1}{\pi} \ln\left(\frac{1}{1 - \sqrt{1 - e^{-2TR/T_1}}}\right) \qquad [8]$$

which is fulfilled for typical imaging applications. In the example discussed above, it holds true for an FOV exceeding 54 mm (together with a $FWHM_{pe}$ of 2 mm) or for an $FWHM_{pe}$ lower than 5.9 mm (together with a FOV of 160 mm). For T_1 values shorter than 1000 ms, such as for white matter or generally at lower field strengths, the limitations are even less restrictive.

Variable Flip Angles

The application of variable flip angles α_i which increase during sequence progression to $\alpha_N = 90°$ (N: number of echoes) promises to fully exploit the available longitudinal magnetization. If M_i denotes the actual longitudinal magnetization for echo number i, its signal S_i is given by $S_i = M_i \cdot \sin \alpha_i$. Because $M_{i+1} = M_i \cdot \cos \alpha_i \cdot \exp(-TR/T_1)$, the signal ratio of successive echoes yields:

$$\frac{S_{i+1}}{S_i} = \frac{\sin \alpha_{i+1}}{\tan \alpha_i} \cdot \exp\left(-\frac{TR}{T1}\right). \qquad [9]$$

To obtain the desired broadness of the PSF, $S_{i+1}/S_i = \exp(-\pi FWHM_{pe}/FOV_{pe})$ must be fulfilled, resulting in:

$$\tan \alpha_i = \frac{1}{\sin \alpha_{i+1} F} \cdot \exp\left(\pi \frac{FWHM_{pe}}{FOV_{pe}} - \frac{TR}{T_1}\right) \qquad [10]$$

that can be used to calculate α_i recursively.

For the half-Fourier example with improved bandwidth (TR = 10.6 ms, α = 13.6°, 44 echoes), the flip angle for the first echo α_1 can be increased to 14.2°, which corresponds to only a modest signal gain of about 4%.

METHODS

Experiments were performed at 2.0 T (Magnetom Vision, Siemens, Erlangen, Germany) using 25 mT m^{-1} gradients (maximum slew rate 40 mT m^{-1} ms^{-1}) and the standard head coil. Written informed consent was obtained from all subjects prior to the examination. Due to software limitations, a fixed flip angle had to be used.

For phantom studies full-Fourier single-shot STEAM MRI (96 echoes) involved an image pixel bandwidth of BW = 560 Hz (TR = 4.3 ms, TE = 3.8 ms) covering an FOV of 192 × 256 mm^2 within an acquisition time of 424 ms (96 × 4.3 ms + 11 ms preparation). Corresponding half-Fourier sequences (52 echoes) either employed the same TR, which decreased the acquisition time to 232 ms (52 × 4.3 ms + 11 ms preparation), or reduced the image pixel bandwidth to BW = 190 Hz (TR = 7.8 ms, TE = 7.3 ms). In close analogy, human studies were performed with full-Fourier phase encoding (80 echoes) at an image pixel bandwidth of BW = 280 Hz (TR = 6.0 ms, TE = 5.6 ms) covering an FOV of 160 × 256 mm^2 within 492 ms (80 × 6.0 ms + 12 ms preparation). Half-Fourier versions (44 echoes) reduced the acquisition time to 276 ms (44 × 6.0 ms + 12 ms preparation) or the image pixel bandwidth to BW = 120 Hz (TR = 10.8 ms, TE = 10.4 ms). For all measurements the in-plane resolution was 2 × 2 mm^2 with a section thickness of 5 mm. The flip angle calculations were performed for a desired $FWHM_{pe}$ of 2 mm assuming a T_1 of 300 ms for the phantom and 1000 ms for human brain studies.

For SNR determinations the mean signal intensity was divided by the SD of the signal σ, which was calculated from the SD of the pixel intensities in a background region taking the different reconstruction methods into account: for full-Fourier phase encoding (magnitude reconstruction) $\sigma = \sqrt{\frac{2}{4} - \pi}\sigma_{FF}$ and for half-Fourier phase encoding (real reconstruction) $\sigma = \sqrt{\frac{\pi}{\pi - 2}}\sigma_{HF}$.

RESULTS AND DISCUSSION

Figure 3 shows single-shot STEAM MR images of a doped water phantom together with the corresponding signal decay in k-space acquired in the absence of a phase encoding gradient. In order to achieve a PSF with an $FWHM_{pe}$ of 2 mm, full-Fourier phase encoding requires a very low flip angle of 3.6° because of the short T_1 (Fig. 3a). A deliberate increase of the flip angle to 11.0° certainly improves the SNR but also causes severe image blurring along the phase encoding dimension of the image (Fig. 3b). In contrast, half-Fourier phase encoding with a flip angle of 11.0° (Fig. 3c) almost doubles the SNR (relative to Fig. 3a) in almost half the acquisition time and without sacrificing the resolution. The use of a reduced bandwidth in conjunction with a properly adjusted flip angle does not result in a further SNR improvement for the short T_1 water phantom (Fig. 3d).

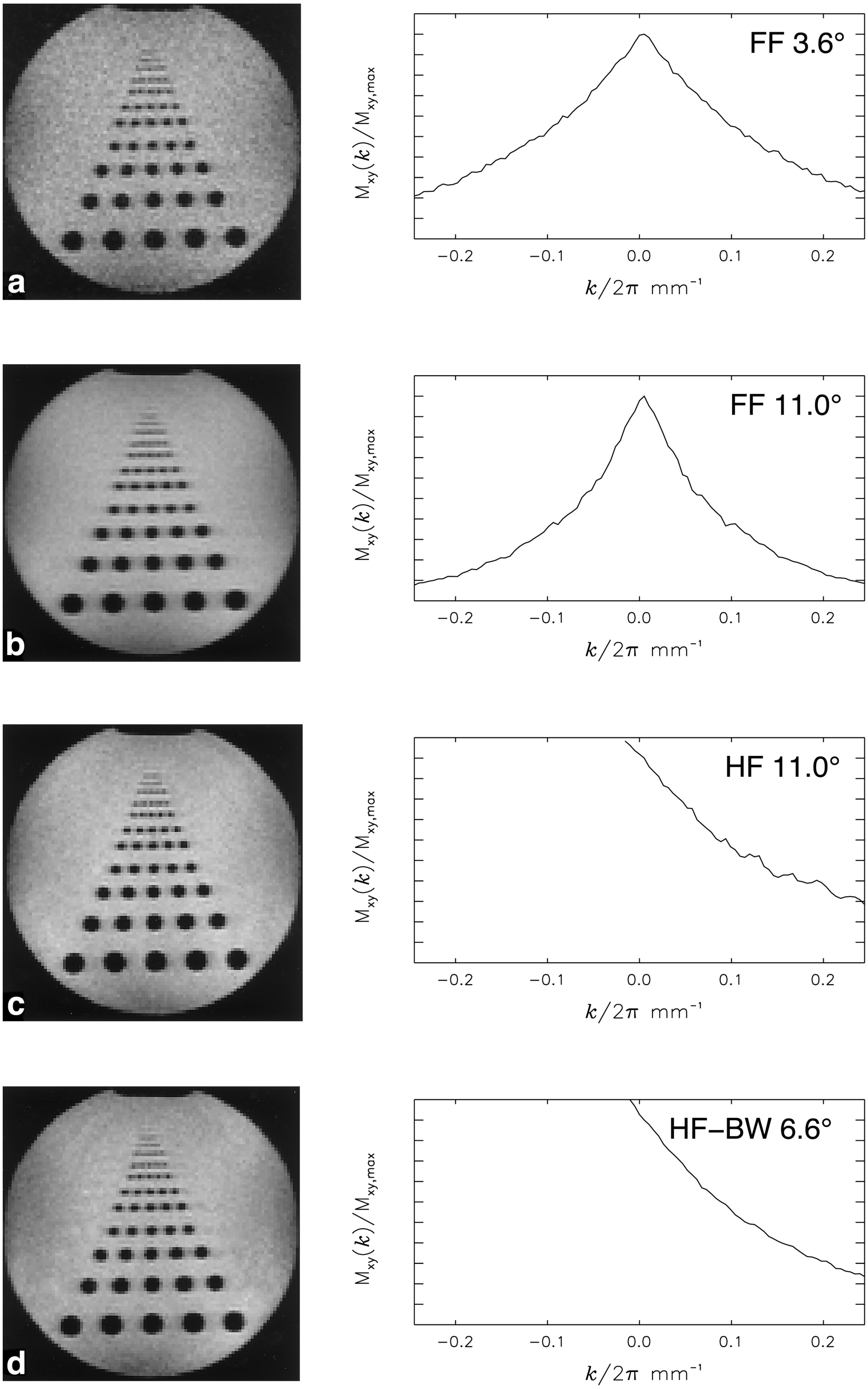

FIG. 3. Single-shot STEAM MR images (left) and experimental signal decay (right) of a doped water phantom for full-Fourier (FF) phase encoding with a flip angle of (**a**) 3.6° and (**b**) 11.0° (acquisition time 424 ms each) as well as for half-Fourier (HF) phase encoding with a flip angle of (**c**) 11.0° (232 ms) and (**d**) 6.6° in conjunction with a reduced image pixel bandwidth (BW, 424 ms).

Figure 4 summarizes corresponding results for single-shot STEAM MRI of the human brain. For the full-Fourier version an increase of the flip angle from 9.4° (Fig. 4a) to 14.7° (Fig. 4b) violates the restriction set by the desired PSF and again results in a higher SNR at the expense of a blurred image. In contrast, half-Fourier phase encoding with the same flip angle of 14.7° (Fig. 4c) does not affect the image resolution, reduces the acquisition time from 492 ms to 276 ms, and maintains the SNR (gray and white matter) as for full-Fourier phase encoding at similar resolution (Fig. 4a). When combining half-Fourier phase encoding with a reduced receiver bandwidth and an adjusted flip angle of 13.6°, the resulting SNR is improved by about 40% (Fig. 4d). The fact that this significant gain is slightly

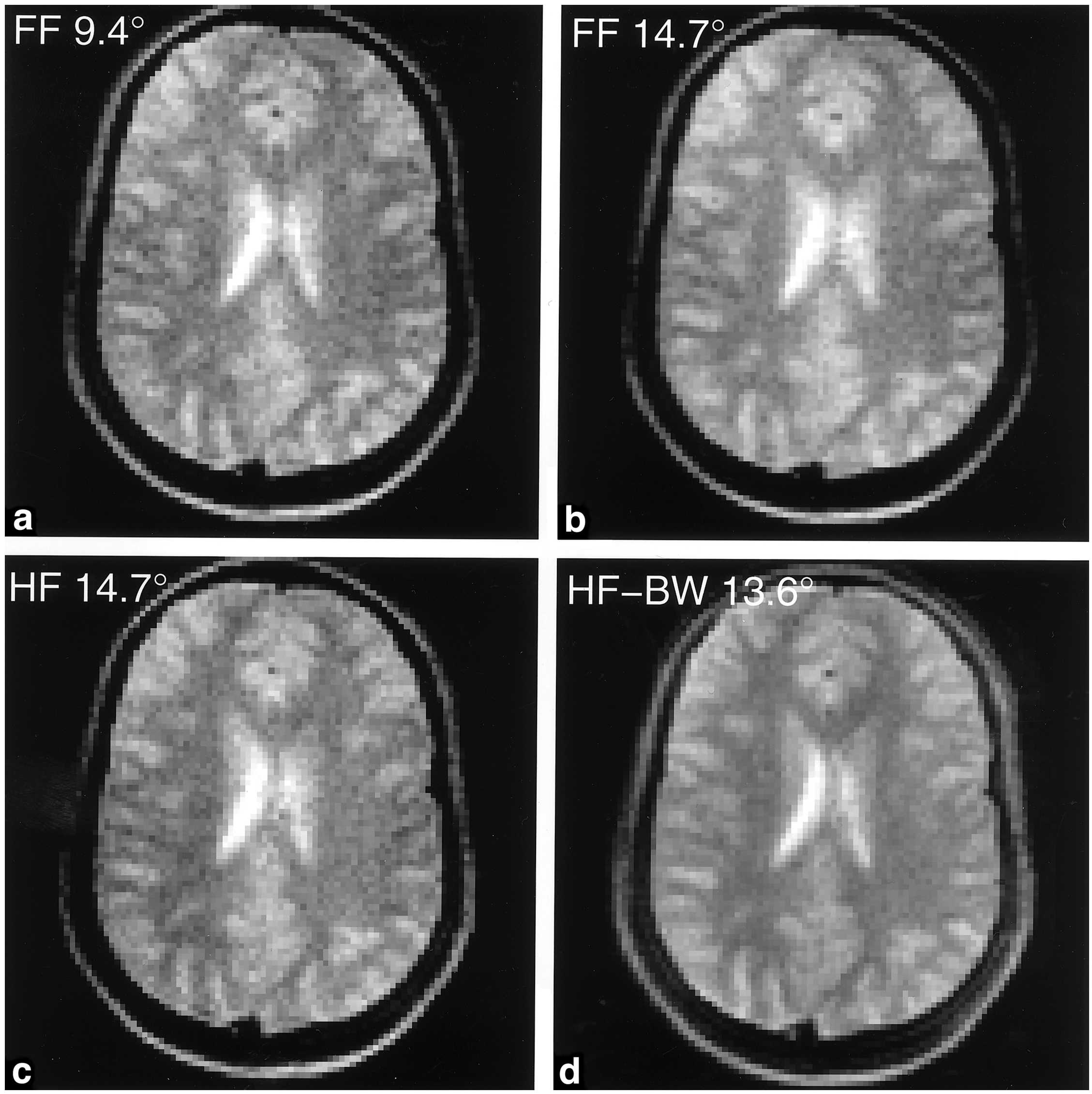

FIG. 4. Single-shot STEAM MR images of the brain of a normal subject for full-Fourier (FF) phase encoding with a flip angle of (**a**) 9.4° and (**b**) 14.7° (acquisition time 492 ms each) as well as for half-Fourier (HF) phase encoding with a flip angle of (**c**) 14.7° (278 ms) and (**d**) 13.6° in conjunction with a reduced image pixel bandwidth (BW, 492 ms).

lower than the expected 50% increase in SNR may be a consequence of neglecting the need for additional central Fourier lines for phase correction in the theoretical derivation.

CONCLUSIONS

In conclusion, half-Fourier phase encoding emerges as an excellent option for single-shot STEAM MRI, which further reduces the measurement time or—perhaps more importantly—significantly increases the available SNR. The approach is expected to enhance the applicability of single-shot STEAM MRI as a high-speed alternative to EPI in areas such as diffusion-weighted imaging, especially in high-field applications.

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