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Multiple-frequency excitation wideband MRI (ME-WMRI)

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Purpose: In this study multiple-frequency excitation wideband MRI (ME-WMRI), a technique designed to acquire simultaneously excited images without blurring was proposed.

Methods: ME-WMRI consists of (a) a coherent acquisition sequence that applies refocusing gradient during readout to mitigate signal dephasing, and (b) a signal enhancement procedure that further reduces image blurring. This two-step method was tested on phantom imaging and in vivo imaging with up to 3.84 pixel blur.

Results: Imaging results in both phantom and in vivo show that the lost details were successfully restored in the blur-mitigated images, and the accelerated images were comparable to standard images.

Conclusions: The proposed ME-WMRI method can effectively undo the image blurring, and thus removes the limitation on fast simultaneous multislice MRI systems with a large acceleration factor.

Key words: ME-WMRI, dephasing, blur, coherent acquisition, signal enhancement

1. INTRODUCTION

Magnetic resonance imaging has become a powerful tool and the fastest growing medical imaging technique since it was first invented four decades ago.1 The rich information and contrast it provides has a profound impact on physicians, biochemists, neurologists, oncologists, psychologists, and many others. Over the past few decades, MR imaging speed has been greatly improved with various acceleration techniques such as parallel imaging and non-Cartesian sampling strategies.2–4 Nevertheless, with the increasing need for large volume coverage scans, we believe the overall MRI throughput could be further improved with simultaneous multislice methods comparable to mlicut method in CT.

The use of simultaneous multislice (SMS) excitation to accelerate MR scans was first brought up more than two decades ago.8 In this work, multiple-frequency signals spread over a bandwidth wider than conventional MRI (wideband MRI) was used in the excitation to increase acquisition efficiency. Such technique can excite different locations of the subject simultaneously, as shown in Fig. 1. Since then, SMS technique has gathered attention and several new methods have been proposed. These methods are categorized into two classes, shown in Fig. 2. The first class uses additional hardware. Multicoil array technique was proposed by Larkman et al., where spatial encoding information in the multicoil receiver system provides information to separate the simultaneously excited slices.9 Another method reported by Breuer et al. implements phase-modulated RF pulses and controlled aliasing to perform multislice imaging with phased array coils.10 Applications such as functional and diffusion imaging have greatly benefited from these techniques, making the imaging of brain faster than ever.11–13 Other than adding RF coils, Lee et al. presented a method that superimposes a micro-B0 array onto the main field to assign unique resonance frequencies to different slices (MAMBA), thus separating the simultaneously excited images.14–16 Setsompop et al. combined CAIPIRINHA technique with multislice echo planar imaging (EPI), and skillfully utilized blipped separation/refocusing gradients to prevent signal dephasing.17,18

On the other hand, there are approaches that achieve multislice imaging without extra hardware. In 1988 Weaver proposed a method that applies a gradient during spatial encoding to modulate the carrier frequency for each excited slice such that they become separated in the reconstructed image.8 Paley et al. combined SENSE imaging technique to further increase the scanning speed of Weaver’s method.19 However, the separation gradient causes dephasing along the slice direction and leads to voxel tilting/in-plane image blurring. Wu et al., the authors of this study, have extended this method to 3D imaging sequences and achieved good results in isotropic brain imaging.20

In addition to the above two classes of SMS methods, Souza et al. reported the use of Hadamard-encoded RF pulses as a means to separate the simultaneously excited/acquired slices during image reconstruction. Similarly, phase-offset multiplanar (POMP) technique proposed by Glover also applies an encoded RF pulse for excitation.21,22 However, these
two methods do not shorten the scan time of MRI due to the extra number of excitations or phase encoding steps required in their implementation.

Although much improvement has been obtained in the previous works, the solution for multislice acquisition of high-quality MR images in commonly used gradient echo or spin echo imaging without the use of added hardware still remains elusive. In this study we start from the work of Weaver and investigate the signal processing aspect of SMS MRI. The underlying mechanism for image blurring is formulated and two new techniques that alleviate the image blurring of both 2D and 3D wideband MR imaging are proposed. Experimental results show that the proposed solution can increase the imaging speed by several folds without compromising the acquired image quality.

2. METHODS

2.A. The effects of separation gradient

According to MR physics, separation gradients applied during readout cause signal dephasing thus image blurring. Without loss of generality, we consider only the signal corresponding to a single excited slice in a SMS MRI system. Although derived with a single-slice scenario, the results can be extended to SMS excitation cases under the assumption of ideal field homogeneity, gradient linearity, and RF coil performances.

The 2D $k$-space (spatial frequency space $k_x$ and $k_y$) MRI signal $S(k_x, k_y)$ derived from Eq. (1), where $z_1$ and $z_2$ represent the two ends of the excited slice, and $\gamma$ is the gyromagnetic ratio. We also assume ideal RF excitation and uniform proton density distribution across the excited slice $\rho(x, y)$:

$$S(k_x, k_y) = \int_{z_1}^{z_2} \left[ \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \rho(x, y) \exp[\iota 2\pi (k_x x + k_y y)] dx dy \right] dz$$

$$= (z_1 - z_2) \times \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \rho(x, y) \exp[\iota \gamma (G_x x \tau + G_y y T_{pe})] dx dy.$$  (1)

The proton density distribution function is subjected to a 2D-spatial encoding process [spatial encoding gradients and duration ($G_x$,$\tau$) and ($G_y$,$T_{pe}$)] and then it is integrated along $X$ and $Y$ directions. The resulting MR image can be obtained through two-dimensional inverse Fourier transform of the sampled $k$-space signal $S(k_x, k_y)$.

In SMS methods after slice excitation, an additional separation gradient was applied during spatial encoding and this separation gradient ($G_z$) has the same duration $\tau$ as the
FIG. 3. Pulse sequence of wideband MRI and the effects in signal magnitude and phase caused by separation gradient. (a) Pulse sequence of 2D wideband MRI with a separation gradient $G_z$ added during frequency encoding. The duration of the separation gradient is the same as frequency encoding gradient and the gradient strength is given that the excited images can be fully separated after reconstruction. (b) The effects on signal magnitude and phase with the application of separation gradient. Due to the phase difference of the spins across the excited slice (green and red arrows at the bottom), the collective magnitude will undergo an attenuation with a sinc-shaped profile along sampling time. There is also a linear phase change that will result in image shift. (c) The normalized signal magnitude and phase in the presence of a separation gradient measured by experiment (1–1.8 pixel blur).

The 2D $k$-space MRI signal is then given by

$$S'(k_x, k_y) = \int_{z_1}^{z_2} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \rho(x, y) \exp\{i2\pi(k_x x + k_y y)\} dx dy dz$$

$$= \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \rho(x, y) \exp\{i\gamma(G_{xx} x \tau + G_{yy} y T_{pe})\} dx dy \times \left( \int_{z_1}^{z_2} \exp(i\gamma G_{zz} z \tau) dz \right).$$

Equation (2) indicates that the extra separation gradient $G_z$ introduces a term that is an integration over the slice thickness along the Z direction, namely, from $z_1$ to $z_2$, and

$$\left( \int_{z_1}^{z_2} \exp(i\gamma G_{zz} z \tau) dz \right) = d \times \text{sinc}(\gamma G_{zz} d/2) \times \exp[i\gamma G_{zz} Z_{cen}].$$

where $d = z_2 - z_1$ and $Z_{cen} = (z_1 + z_2)/2$. The MRI signal can be reformulated as

$$S'(k_x, k_y) = S(k_x, k_y) \times \text{sinc}(\alpha k_x) \times \exp(i\beta k_x),$$

where $\alpha = \pi R d$, $\beta = 2\pi R Z_{cen}$, and $R = G_z/G_{x}$.

The sinc term in Eq. (4) causes signal attenuation in higher spatial frequencies ($k_x$), which blurs the image along the X dimension. The degree of attenuation is determined by gradient ratio $R$, duration of the separation gradient $\tau$, and the excited slice thickness $d$. Degree of blur is defined in terms of pixels, in correspondence to voxel tilting in prior studies. From the attenuation profile, the number of pixel blur is also equivalent to the ratio of acquisition length to the width of the first zero crossing of the sinc function.

The exponential term in Eq. (4) represents a shift along the X direction in the acquired image, and the amount of shift $x_s$ turns out to be the same amount as $R Z_{cen}$, where $K$ is the slope of the tilted voxels defined in previous SMA studies and $Z_{cen}$ is the distance between the slice and the center of the separation gradient. The sequences and the effects of the separation gradient during readout derived in Eq. (4) are shown in Fig. 3. Shown in the far right of Fig. 3 is the acquired attenuation profiles from 1 to 1.8 pixel blur to verify the equations.

At 1 pixel blur the span of the first zero crossing is about the total acquisition length. Note how the span decreases as the blurring becomes more serious.

For the sake of convenience, we define the number of simultaneous excited slices in ME-WMRI as the “wideband factor” and use $W$ to represent this factor in the following discussion. During acquisition, the FOV along readout as to be expanded $W$ folds to contain the separated slices, number of sampling along readout $N_{FE}$ was also increased by $W$ times to maintain the same spatial resolution.

**2.B. Coherent acquisition method**

After identifying the source of image blurring, we propose a method that diminishes signal dephasing in the acquired $k$-space data. This method breaks one readout duration into several segments separated by refocusing gradients. As a result, the signal during each segment will experience less signal attenuation and the final attenuation profile will become cycloid shaped with a series of sinc lobes. We call...
this method coherent acquisition method, and the number of segments per readout duration is defined as “S.” Therefore, the conventional SMS MRI corresponds to the $S = 1$ case.

In Fig. 4(a), we show an $S = 2$ sequence and the evolution of the effect on the signal over time, magnitude, and phase drawn separately. The sequence depicts the actual gradient waveform with finite slew rate, buffer points were added at the guard-time intervals (shaded areas) right before and after refocusing to ensure equally spaced sampling between the segments, also no readout gradient was applied during the refocusing period. The magnitude and duration of the refocusing gradient is designed to minimize signal dephasing in each segment. In the $S = 2$ sequence since MR signal is sampled during the refocusing period, redundant information such as $k$-data acquired during refocusing, unequally spaced samplings, and overlapping $k_z$ will be removed and the remaining $k$-space data segments joined together before further signal processing.

2.C. Signal enhancement method

Even after coherent acquisition sequence, there still remains some residual $k$-space signal attenuation located in the valleys of the cycloid shaped profile. The degree of blurring was decreased for certain structures, and the Fourier transform of this attenuation profile introduced smaller spikes that would cause additional image artifacts. Toward this end, we propose to further apply $k$-space signal enhancement to restore the original signal intensity.\(^{25}\) Considering the image property and attenuation level we choose Wiener filtering, which is given by \( \hat{K}(u, v) = \frac{1}{H(u, v) \left| H(u, v) \right|^2 + P(u, v)} \) \( K(u, v), \) (5)

where the acquired $k$-space data are $K(u, v)$ and the attenuation curve $H(u, v)$. The attenuation curve $H(u, v)$ can either be measured or simulated. The advantage of Wiener filter over a direct inverse filtering is that by adding a weighting term $P(u, v)$ ($P = 1/$SNR), the Wiener filtering is able to adjust the level of signal enhancement in images of low SNR, avoiding unwanted noise amplification. A $P$-weighting map is calculated over the acquired $k$-space to determine the level of enhancement at different spatial frequencies. The process is described in Fig. 4(b).

2.D. Experiment setup

The data obtained for this study are performed on a Bruker 3T Medspec system using a T8826 transmit/receive head coil as well as a 7 T Biospec animal system using a volume coil. Images of phantom made by Bruker for quality testing are acquired to evaluate the effectiveness of the blur mitigation method. The $S = 2$ images have a FOV of $19 \times 19$ cm, flip angle (FA) = 30°, TR/TE = 100/8 ms, slice thickness = 1.5 mm, resolution = 0.781 $\times$ 0.781 mm, echo position was set at 30%, falling on the center of the first readout segment and away from the gradient switching. Separation gradient strength is set so that the image blur ranges from 0 to 2 pixels, in steps of 0.2 pixel. Gradient strength = 70% of
FIG. 5. Effects of the two-stage blur mitigation methods and SNR analysis. Contrasts of low-resolution detail (L), high-resolution detail (H), and signal-to-noise ratio are plotted with respect to different degrees of blurring. After applying blur mitigation methods, the detail contrast of both frequencies remains on par with standard images up to 1.8 pixel blur, but the low resolution detail contrasts start to degrade over 1.8 pixels. SNR of all degrees of blur remains the same due to the adaptive Wiener filtering that prevents noise from being enhanced.

3.02 G/cm, ramp time = 450 μs, slew rate = 4.69 mT/m ms. Attenuation profiles corresponding to various degrees of blur are also taken to verify the equations and to perform signal enhancement. The profiles are normalized by the free induction decay baseline to derive the actual attenuation. To quantify the effectiveness of the blur mitigation methods, the image profiles over vertical stripes of high- and low-spatial frequencies at the lower part of the phantom are taken. Detail contrast is defined as the average peak-to-valley over mean signal strength, and will decrease as the image becomes blurrier. As for the W = 2, S = 3 experiment a custom-made phantom is built to show blur mitigation effects in fine structures. Two slices 3 cm apart are excited by a 51 kHz modulated sinc pulse and the power is adjusted to maintain the equivalent flip angle (FA = 30°). The W = 2 images have a FOV = 12 × 6 cm, TR/TE = 100/9 (ms), a 1.28 ms increase in TE was caused by the added guard points. Gradient strength = 15% of 3 G/cm, ramp time = 450 μs/slew rate = 1 mT/m ms. Echo position was set at 50%, to appear at the center of the second read-out segment. The in-plane resolution = 0.3125 mm and slice thickness = 0.6 mm, and the image is blurred 3.84 pixels. Signal and noise were measured, respectively, to derive the SNR before and after applying blur mitigation methods in all phantom experiments. Noise was measure from the corners of the image and signal as the average of various ROIs, as shown in Fig. 5.

Other than phantom, a W = 2, S = 3 pig’s tail image is taken to demonstrate the capability of blur mitigation in in vivo scans. The in vivo image has a 3 × 3 cm FOV, 0.156 × 0.156 mm in-plane resolution, flip angle = 30, TR/TE = 100/9 ms, also a 1.28 ms increase in TE. Gradient strength = 20% of 3 G/cm, ramp time = 450 μs, slew rate = 2.66 mT/m ms, and 0.6 mm slice thickness. The two excited slices were 45 mm apart, resulting in a 2.56 pixel blur.

As for 3D implementation of ME-WMRI, the human brain imaging has a W factor of 3, 19.2 × 19.2 × 25.6 cm³ FOV, 0.75 mm isotropic resolution, flip angle = 25°, TR/TE = 30/6 ms gradient strength = 70% of 3 G/cm, ramp time = 450 μs, slew rate = 4.66 mT/m ms. The W = 6 mice whole body image has a 3 × 2.5 × 9 cm³ FOV, 0.156 mm isotropic resolution, flip angle = 30°, TR/TE = 30/7 ms gradient strength = 50% of 3 G/cm, ramp time = 450 μs, slew rate = 2.66 mT/m ms.

3. RESULTS

The quantitative results of blur mitigation are illustrated in Fig. 5. The two plots on the right display detail contrast and SNR before and after applying the blur mitigation methods. The high- and low-frequency detail contrasts of the standard image are represented by the red and black horizontal lines on the bottom-right plot. Successful blur mitigation is achieved when the detail contrasts approach these two lines, respectively.

3.A. Phantom imaging

3.A.1. S = 1 image with signal enhancement

In S = 1 imaging the difference between the standard image and blurred images becomes obvious when the blur exceeds 1 pixel, both the high and low frequency stripes in the
phantom are blurred by the separation gradient (S1 L and S1 H). Having known the attenuation profile for each degree of blur we apply signal enhancement to the acquired $S = 1$ k-space. After signal enhancement, the high resolution details increase ($S_1$ enhanced H) while the low resolution details remain the same ($S_1$ enhanced L).

3.A.2. $S = 2$ blur mitigated images

It is clear from the start that the $S = 2$ images have better detail contrast in high spatial frequency ($S_2$ joined H) than the $S = 1$ blurred images ($S_1$ H) up to 2 pixel blur. However, the low frequency contrast did not improve for any degree of blur ($S_2$ joined L). Such result indicates that in $S = 2$ coherent acquisition, high spatial frequency signals are recovered while low frequency signals still suffer attenuation. The detail contrast of high spatial frequency drops to almost zero for images with a blurring greater than 1.8 pixels.

By enhancing the $S = 2$ acquired image signals, both low- and high-resolution contrasts are further improved ($S_2$ enhanced L and $S_2$ enhanced H). The detail contrasts of low spatial frequency in $S = 2$ enhanced images ($S_2$ enhanced L) have the equivalent results as the $S = 1$ enhanced images ($S_1$ enhanced L), while the high spatial frequency details were improved across all degrees of blur and become comparable to the standard image.

3.A.3. $W = 2$ blur mitigated phantom images

Figure 6 shows the $W = 2$ accelerated ME-WMRI with $S = 3$ sequence and signal enhancement applied. The comparison images are displayed to show the effectiveness of the blur...
mitigation methods. The measured image profiles of vertical stripes show noticeable improvement after applying blur mitigation, recovering most of the peak-to-valley contrast. This result indicates the use of a higher $S$ can recover image detail from a more severe degree of blur incurred by acceleration with large $W$ acceleration or demanding geometrical setting. We have also measured the signal and noise of the image, respectively, and derived SNR for each slice during each blur mitigation step.

3.B. In vivo imaging results

Figure 7 shows the potential of blur mitigation methods for $W = 2, S = 3$ in vivo imaging. The axial images of a pig’s tail were blurred 2.56 pixels, causing boundaries and structures to be soft and indiscernible. With the blur mitigation methods applied, the contours of tissue and cartilage maintain their sharpness, which undoubtedly helps to improve the diagnosis accuracy.

3.C. 3D implementation

A straightforward extension of the proposed methods is Wideband MRI in 3D, where the equivalent thickness can be easily decreased to produce sharper images through adding the number of phase encoding steps along slab thickness. Shown in Fig. 8 are the sequence and results of 3D wideband MRI. With a thinner image slice, 3D wideband MRI can be acquired with image blur that can be neglected. One concern is the modulated RF pulse for multiple slab excitation may contain wider transitional bands between the excited slabs and result in image dark bands.

4. DISCUSSION

4.A. Image quality: Detail contrast and SNR

In the $S = 1$ phantom image high-frequency detail contrast drops rapidly and reaches bottom at 1.8 pixel blur while low-frequency detail contrast maintains approximately the same value up to 1 pixel blur. After enhancing the $S = 1$ image, it is obvious that the low-frequency detail contrasts have greatly improved but the high-frequency contrasts do not change much. That is due to the attenuation at high frequency components cannot be recovered after being strongly attenuated. On the other hand the $S = 2$ images show a much better high-frequency contrast even before enhancement, but the low-frequency contrasts are on par with the unenhanced $S = 1$ images. Both low- and high-frequency contrasts of the $S = 2$ images are improved after applying Wiener filter signal enhancement. The low-frequency contrasts slowly decay and the high-frequency contrasts stay at a relatively high level throughout all sets. In this case, the two blur mitigation methods can deal with structures of different spatial frequencies in the phantom. It is conceivable that for images without very sharp details, signal enhancement alone is sufficient to mitigate the blur. One can also design the attenuation profile of the coherent acquisition scheme to retrieve desired frequency components by adjusting the refocusing gradients and length of each readout segments.
FIG. 8. 3D ME-WMRI sequence and $W = 3$ isotropic brain and $W = 6$ mice whole body imaging. (a) In 3D ME-WMRI, the equivalent image thickness can be easily made thinner with the cost of more $z$ encoding steps. Accelerated isotropic resolution images can be acquired with minimal blurring. (b) A $W = 3$, 0.75 mm isotropic whole brain image was taken under 10 min of scan time whereas a standard scan will take about half an hour without other acceleration techniques. (c) Another example is a $W = 6$ isotropic mice whole body imaging, the imaging time was reduced from 1 h to less than 10 min, demonstrating the advantage of the proposed ME-WMRI imaging scheme for scanning long objects.

On the subject of image contrast, the addition of refocusing period leads to a slightly longer acquisition duration and a longer TE if a greater receiving bandwidth was not used. Segments of the readout will have different effective TE’s but the final image contrast will be dominated by the effective TE where the image echo locates. In the $W = 2$, $S = 3$ experiment the TE of image echo is prolonged by 1.28 ms, explaining the minor difference in image contrast. This difference in TE could be minimized by optimized sequence design.

Other than longer TE, the additional separation/refocusing gradients could cause differences in echo intensity thus affecting SNR. In the $S = 2$ situation where the image echo is located in the first segment, the additional gradient switching generated a refocusing effect causing the echo to have a slightly higher value ($\sim$5% increase compared to standard image). However, in the $S = 3$ experiments, image echo that was acquired in the second segment shows some decrease (the joined $k$-space has a $\sim$8% decrease compared to standard image, but was later restored by Wiener filtering). This effect is mainly governed by the field homogeneity, gradient performance, and sequence design; fine shimming and careful gradient adjustments could minimize the difference.

Since the number of frequency encoding increases along with the receiving bandwidth (both $W$ times increase), noise per pixel in ME-WMRI remains unchanged from the conventional MRI. Therefore, the SNR per pixel in ME-WMRI should remain approximately the same for all $W$ factors given equivalent excitation/reception.26

4.B. Field homogeneity and gradient performance

For an ideal gradient the switching process in coherent acquisition sequence should be instantaneous and therefore the resonance signal peaks caused by the refocusing effect should appear precisely at the designated time instants. However, in real-world hardware the gradients have limited slew rates and cannot be switched instantaneously, the ramped gradients and induced Eddy current could cause errors in readout sample values. We have inserted a few dummy sample points (a short guard-time interval) in our sequences to serve as a buffer between each gradient switching and prevent $k$-data discontinuities. There will be an overlapping in the $k$-space trajectory that can be either discarded or used for gridding during image reconstruction. The guard-interval duration depends on the capability of the gradient coils in the MR scanners. For sequences that use greater gradient strengths, shorter rise time, and less guard points, the accurate knowledge of $k$-space trajectory might be required for reconstruction.

As for clinical considerations where the switching of gradients might cause stimulation of peripheral nervous systems, the trapezoid gradient waveforms used in this study produced a maximum slew rate of 6.5 mT/m ms, which is well under the PNS stimulation threshold.27

As for transmission and reception, the use of multichanneled coils and algorithm designed pulses will increase the excitation uniformity among slices,28–31 while traveling wave technique could provide unvarying reception quality across a larger coverage.32

Other than hardware limitation, the decay of FID is inevitable across the acquisition period. Major factors such as field homogeneity, substance composition, etc., will determine the speed of decay. Such decay causes discontinuity in the $k$-space data among the acquired segments and may very well induce image artifacts. Precise shimming at the locations of multiple-slices is suggested to avoid such issue.
4.C. Advanced sequence design and implementation

The overall effectiveness of the proposed method was contributed by k-space signal distribution and the multisegmented sequence design. In general, the higher the number of segments, the less attenuation of the k-space signals and the lower distortion in enhanced images. A higher S factor will be needed to refocus the rapidly attenuating signals in scans with higher W factor. Locations and lengths of the segments could also be modified according to the signal distribution of k-space to optimize blur mitigation. Considering signal enhancement, the sinc attenuation profiles of a magnetic field of good linearity would be highly identical to the simulation results, meaning that there is no need for attenuation profile acquisition in each scan.

As for advanced implementations of coherent acquisition sequences, the readout segments need not to be equally spaced along the acquisition. One can choose specific regions of the image k-space and adjust the gradients accordingly to preserve the most meaningful frequency components from being attenuated. Furthermore, a smaller S factor will be enough with schemes that employ shorter readout durations such as partial k-space methods. In combination with compressed sensing, the peaks of severely attenuated profiles can be placed at the sparsely sampled points, obtaining fine k-space data with high W acceleration factors. SMS imaging was put together with other methods to achieve a higher acceleration rate in clinical studies, 10–12 parallel imaging and other acceleration techniques could be added onto ME-WMRI to further increase the scanning throughput as well.33

Pushing coherent acquisition sequence to the limits is a set of fast switching separation and readout gradients with refocusing between each sampling point to prevent dephasing. There are two advantages in this design: at first, the total acquisition time can remain unchanged since no additional time is spent on refocusing; and second the required magnitude of refocusing gradient would decrease.

5. CONCLUSIONS

The proposed ME-WMRI method can effectively undo the image blurring caused by the separation gradient during acceleration. Coherent acquisition sequence alleviates signal dephasing by breaking a continuous readout into multiple segments, while signal enhancement further restores the attenuated frequency components. With the two proposed methods, ME-WMRI can successfully mitigate the image blur in both phantom and in vivo imaging, and thus removes the limitation on fast SMS MRI systems with a large W factor.

With the image blurring solved by the proposed methods, the high temporal/spatial resolution capability of ME-WMRI could provide a new MRI experience for practitioners and patients alike. We believe that ME-WMRI provides new possibilities for future generations of MRI, in which increased temporal/spatial resolution contributes greatly to whole body screening, early cancer detection, and other new MRI applications.34–37

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