Turbo Fast Three-Dimensional Carotid Artery Black-Blood MRI by Combining Three-Dimensional MERGE Sequence with Compressed Sensing

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**INTRODUCTION**

A three-dimensional (3D) motion-sensitizing driven equilibrium (MSDE) rapid gradient echo (3D MERGE) technique has been demonstrated as a useful tool for in vivo assessment of atherosclerotic disease in 3D black-blood MRI (1). It can accurately depict plaque in all three spatial directions and thus provide more accurate measurement of plaque morphology, improve the reproducibility of artery MRI, and allow registration of images in serial studies (2,3). However, a relatively long scan time for 3D phase encoding is needed to sufficiently cover image volume and to achieve satisfactory spatial resolution (4). In particular, the frequent repetition of black-blood preparation ahead of the imaging acquisition further decreases scan efficiency. As a result of long acquisition time, image quality is more likely to be negatively affected by motion such as swallowing, respiration and neck movements (5–7). Therefore, a method to accelerate 3D black-blood MRI is highly desirable.

Many efforts have been made for the development of undersampled MR acquisitions to reduce scan time (6–10). Among them, compressed sensing (CS) method has attracted wide attention due to its ability to recover an image from data sampled below the Nyquist frequency without degrading image quality (11–15). In CS, if the underlying imaging exhibits sparsity in the image (pixel) or a transform domain, then the image can be recovered from a randomly undersampled dataset (14). Most MR images are sparse within the finite differences transform, wavelet transform and other transform domains (14).

In this study, we sought to investigate the feasibility of turbo fast 3D black-blood imaging by combining a 3D MERGE sequence with CS (CS-3D MERGE) at a 3 T MR scanner. To effectively suppress flow signal in undersampled k-space, a pseudo-centric phase encoding order was developed for CS-3D MERGE. We quantitatively measured undersampled and fully sampled images and performed statistical analyses on healthy volunteers for in vivo experiments. Moreover, isotropic high resolution images using different CS acceleration factors were acquired to further evaluate the performance of the proposed technique.

**METHODS**

In this article, $k_x$, $k_y$, and $k_z$ represent frequency encoding, in-plane phase encoding and slice phase encoding in a Cartesian 3D k-space.
FIG. 1. Sequence diagram of a 3D MERGE sequence. An MSDE black-blood preparation with a length of TM, consists of a 90°-180°-180°-90° hard RF pulses and motion sensitizing gradients.

3D MERGE Sequence

The 3D MERGE black-blood sequence consists of a 3D spoiled segmented fast low angle shot sequence and an MSDE preparation (1) (Fig. 1). The MSDE black-blood preparation consists of a 90° hard RF pulse, two 180° refocusing hard RF pulses, a 90° hard RF pulse and four motion sensitizing gradients. Within the sequence, spoiler gradients are applied to crush residual transverse magnetization and fat suppression is used to generate optimal delineation of the outer vessel wall. The imaging sequence is a fast low angle shot sequence with centric phase-encoding order (CPEO).

A Pseudo-Centric Phase Encoding Order for Maintaining Black-Blood Contrast in CS-3D MERGE

In a 3D-MERGE sequence, maximal blood suppression occurs immediately following the black-blood preparation. Thus, only a limited amount of phase encoding samples can be collected after preparation. The acquisition order of these phase encoding samples should be arranged in CPEO with a symmetrical low-high order. Therefore, the full 3D k-space is usually separated into several groups or segments to fill k-space in this order. In the undersampled k-space of CS-3D MERGE, the acquisition order of those random samples also needs to be arranged so as to be consistent with the CPEO.

As illustrated in Figure 2, the 3D phase encoding loop structure in a 3D-MERGE sequence is a centric k (slice) loop nested within a sequential k (phase) loop, and the number of acquisitions following one MSDE preparation is equal to the number of slice. In Figure 2, the bold numbers on the left side of the k-space represent the filling sequence in a conventional CPEO for full sampling, and the colored solid circles are the random samples in the undersampled k-space. The colored solid circles linked in a line represent a group of samples acquired after one black-blood preparation.

First, random samples with positions that perfectly match the CPEO along k are grouped into segments (blue and orange groups in Fig. 2). In this case, the encoding number of each segment is equal to the slice number. The numbers in the colored solid circles show the order in which that group is filled. Next, all remaining samples are grouped into segments (green and purple groups). Although their positions do not perfectly match, they approximately coincide with the CPEO. Here, the encoding number of each group is equal to the slice number. In the last group, however, it has less encoding number than the slice number (red group).

3D CS Reconstruction

For 3D MRI, after one-dimensional inverse Fourier transformation along k, the signal s at each x position is obtained and can be expressed as:

\[ s = \Phi m + n. \]

where s is the signal vector, \( \Phi \) is the undersampled Fourier operator obtained by a measurement matrix. A Monte-Carlo method is used to construct the measurement matrix (14). m is the unknown image estimation and n is a vector representing independent and identically distributed (i.d.d) additive white Gaussian noise. Thus, it is probable that m can be exactly reconstructed if m is sparse in a transform domain by solving the following \( \ell^1 \)-norm optimization problem:

\[ \minimize ||\Phi m - s||_2^2 + \lambda_W||\Phi m||_1 + \lambda_{TV} TV(m), \]

where \( \psi \) is the sparsifying transform, \( TV \) is the total variation of m, \( \lambda_W \) is the regularization weight for the sparsifying transform and \( \lambda_{TV} \) is the TV regularization term. The total variation is the \( \ell^1 \)-norm of the finite differences:

\[ TV(x) = \sum_{ij} \sqrt{(x_{i+1,j} - x_{ij})^2 + (x_{ij+1} - x_{ij})^2}, \]

where x is an image matrix.

FIG. 2. Schematic diagram of pseudo-centric phase encoding order. The bold numbers on the left side of the 3D k-space represent the filling sequence in a conventional CPEO, and the colored solid circles are the random samples in the 3D k-space. The blue and orange groups perfectly match the CPEO. The green, purple, and red groups approximately coincide with the centric order. The numbers in the colored solid circles show the order in which that group is filled. The encoding number of every group equals the slice number (red group may have less encodings in the green and purple groups, the encodings on the same row should be acquired completely before collecting next sample.
In this study, Ψ is the identity transform. After one-dimensional inverse Fourier transformation along the frequency encoding, two-dimensional CS reconstruction was performed by solving m with the undersampled k-space data for each coronal slice. The steps of reconstruction are as follows. We first chose an undersampled dataset from one channel of the eight-channel coil for coronal reconstruction. Initial values of λ_W and λ_TV were set to be 0.02. The undersampled datasets were reconstructed by changing λ_W and λ_TV until an acceptable quality for each coronal image was obtained. We then recorded the values of the two parameters and used them for the reconstruction of the other seven channel datasets.

In Vivo Experiments

To assess the performance of the proposed technique of CS-3D MERGE, nine healthy volunteers (age 31 ± 6 years, five males, four females) were recruited. Written informed consent was received from each subject as approved by the local institutional review board. MR images of the carotid arteries were acquired using a clinical 3 T scanner (Signa TM; GE Medical Systems, Milwaukee, WI) with an eight-channel phased-array bilateral carotid coil (GE Medical Systems). The study design consisted of two parts: (1) Part I, statistical comparison between the fully sampled and undersampled images in seven subjects with a spatial resolution of 0.78 × 0.78 × 2 mm³; (2) Part II, qualitative evaluation of isotropic high spatial resolution images for all acceleration factors in two other subjects with a spatial resolution of 0.7 × 0.7 × 0.7 mm³. Part II studies were conducted to further evaluate the performance of the proposed CS-3D MERGE in the presence of high noise level typically observed in high-resolution images.

Using an axial acquisition plane, one fully sampled 3D MERGE (1-fold acceleration) and five CS-3D MERGEs at 2, 2.5, 3, 4, and 5-fold accelerations were acquired in all subjects. For MSDE black-blood preparation, the amplitude, slew rate and duration of the motion-sensitizing gradients pulses were set to 30 mT/m, 120 mT/m/ms, and 2 ms. Total time of the prepulse was 23.1 ms and maximum available gradient strength and slew rate were 40 mT/m and 200 mT/m/ms, respectively.

For part I, the imaging parameters were: pulse repetition time/echo time 6.2/2.9 ms, flip angle 8°, field of view 200 × 200 mm², receiver bandwidth 244 Hz/pixel, matrix size 256 × 256, slice thickness 2 mm, number of slices 32, signal acquisition 1. The spatial resolution was 0.78 × 0.78 × 2 mm³. After zero fill interpolation, it was 0.39 × 0.39 × 2 mm³. The scan time for the six sequential 3D MERGEs was 65, 33, 26, 22, 16, and 13 s, respectively, at 1, 2, 2.5, 3, 4, and 5-fold accelerations.

For part II, the parameters were: pulse repetition time/echo time 6.7/3.0 ms, flip angle 8°, field of view 180 × 180 mm², receiver bandwidth 244 Hz/pixel, matrix size 256 × 256, interpolated to 512 × 512, slice thickness 0.7 mm, number of slices 64, signal acquisition 1. The spatial resolution was 0.7 × 0.7 × 0.7 mm³, and it was zero interpolated to 0.35 × 0.35 × 0.35 mm³. The scan time was 120, 60, 48, 40, 30, and 24 s at 1, 2, 2.5, 3, 4, and 5-fold accelerations, respectively.

Image Analysis

As noise depends on the parameters λ_W and λ_TV in the CS reconstruction algorithm, the lumen signal-to-tissue ratio (STR) was used as a measure of flow suppression efficiency instead of the lumen signal-to-noise ratio. The contrast-to-tissue ratio (CTR) between the carotid arterial lumen and wall was used to evaluate the quality of black-blood imaging.

The STR for lumen was defined by

$$
\text{STR} = 100 \times \frac{S}{T}.
$$

where S is the signal intensity of lumen and T is the signal intensity of the sternocleidomastoid, which is close to the vessel and has a relatively uniform signal intensity. The lumen signal (S_l) was measured as the mean signal obtained from a regions-of-interest drawn in the vessel lumen, while the signal for carotid arterial wall (S_W) was measured as the average value of the manually delineated 1 pixel width path to outline the carotid arterial wall. T was measured as the mean signal intensity within a manually drawn regions-of-interest no smaller than 50 mm² from the sternocleidomastoid. The CTR of wall-lumen was calculated by

$$
\text{CTR} = 100 \times \frac{(S_W - S_L)}{T}.
$$

STR and CTR values were averaged to yield one STR and one CTR for each acceleration. Thus, six sets of STR

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Full sampled dataset (1x)</th>
<th>2x</th>
<th>2.5x</th>
<th>3x</th>
<th>4x</th>
<th>5x</th>
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<td>12.5</td>
<td>16.7</td>
<td>14.4</td>
<td>20.3</td>
</tr>
<tr>
<td>C</td>
<td>20.0</td>
<td>14.5</td>
<td>16.3</td>
<td>17.8</td>
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<td>16.3</td>
</tr>
<tr>
<td>D</td>
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<td>18.3</td>
<td>24.2</td>
<td>26.5</td>
<td>17.9</td>
<td>19.7</td>
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<tr>
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<td>22.1</td>
<td>21.6</td>
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<td>25.0</td>
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<tr>
<td>F</td>
<td>15.4</td>
<td>11.9</td>
<td>12.3</td>
<td>13.8</td>
<td>16.9</td>
<td>18.4</td>
</tr>
<tr>
<td>G</td>
<td>21.3</td>
<td>19.6</td>
<td>20.7</td>
<td>22.6</td>
<td>19.2</td>
<td>22.1</td>
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<td>Mean ± SD</td>
<td>16.67 ± 4.48</td>
<td>15.46 ± 3.73</td>
<td>17.26 ± 5.04</td>
<td>19.13 ± 4.59</td>
<td>17.14 ± 3.54</td>
<td>19.54 ± 3.40</td>
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<td>P value</td>
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<td>0.644</td>
<td>0.105</td>
<td>0.640</td>
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N.A. = not applicable.
and CTR values were obtained for each volunteer. \( T \) was measured from a similar region for all accelerations for each subject to ensure unbiased calculation of STR and CTR values. To fairly compare CTR values with respect to scan time differences, CTR efficiency \( (\text{CTReff}) \) was calculated by the relation:

\[
\text{CTReff} = \frac{\text{CTR}}{\text{SI}_{\text{TH}} \sqrt{\text{TASLICE}}},
\]

where \( \text{SI}_{\text{TH}} \) is the imaging slice thickness (in millimeters) and \( \text{TASLICE} \) is the imaging time per slice (in minutes).

For statistical comparison, the STR, CTR, and CTR\text{eff} were calculated. Two board-certified radiologists with at least 5 years of experience in vascular MR imaging, who were blinded to the technique used, reviewed all images for quality.

### Statistical Analysis

Ten axial images centered at the bifurcation of each carotid artery were used for analysis. Paired \( t \)-tests were performed for both STR and CTR comparison for the in vivo experiments. \( P \) value less than 0.05 was considered significant. Statistical analysis was performed using SPSS (v11.0, SPSS Inc., Chicago, IL).

### RESULTS

The measured values of lumen STR, wall-lumen CTR, and wall-lumen CTR\text{eff} were presented in Figure 3. The mean and standard deviation were 16.67 ± 4.48, 15.46 ± 3.73, 17.26 ± 5.04, 19.13 ± 4.59, 17.14 ± 3.54, and 19.54 ± 3.40 for accelerations of 1x, 2x, 2.5x, 3x, 4x, and 5x in lumen STR, respectively (Table 1). There were no significant differences between the fully sampled and undersampled datasets of 2\(-/5x\). The mean and standard deviation were 29.60 ± 5.28, 32.96 ± 8.90, 30.99 ± 9.36, 29.66 ± 6.31, 28.93 ± 8.00 and 27.36 ± 7.26 for 1\(-/5x\) acceleration acquisitions in wall-lumen CTR, respectively (Table 2). There were no statistically significant differences in wall-lumen CTR between the fully sampled dataset and each undersampled dataset either. For wall-lumen CTR\text{eff}, the mean and standard deviation were 80.56 ± 14.36, 125.69 ± 33.93, 133.65 ± 40.36, 137.90 ± 29.34, 157.47 ± 43.57, and 164.97 ± 43.76 for 1\(-/5x\) accelerated imaging, respectively (Table 3). CTR\text{eff} for accelerations of 2\(-/5x\) were significantly improved when compared to fully sampled imaging (\( P \) < 0.05 for all paired \( t \)-test comparisons).

Figure 4a showed representative images with a resolution of 0.78 \times 0.78 \times 2 \text{ mm}^3 (Part I study). Both radiologists found that all of the undersampled images exhibited comparable flow suppression efficiency, when compared with the fully sampled images. The delineations of outer vessel wall boundary and lumen-wall interface were also comparable with the fully sampled images, except for the slightly blurred vessel wall boundary at 5-fold acceleration. By using 3D multi planar reformating, the oblique reformatted images at accelerations of 2\(-/5x\) exhibited comparable image quality to full acquisition, as demonstrated in Figure 4b. For all undersampled images, comparable image quality at the carotid
bifurcation, where atherosclerosis is usually caused by plaque deposition, were obtained. Figure 5 depicted images from isotropic high spatial resolution imaging. In Figure 5a, there were three representative axial slices. Each radiologist concluded that all undersampled images exhibited comparable flow suppression efficiency and delineation of outer vessel wall boundary and lumen-wall interface with the fully sampled images. In the oblique reformatted images, image quality at the carotid bifurcation was comparable between undersampled and fully sampled images (Fig. 5b).

Table 3

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Full sampled dataset (1x)</th>
<th>2x</th>
<th>2.5x</th>
<th>3x</th>
<th>4x</th>
<th>5x</th>
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<td>178.0</td>
<td>176.7</td>
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<tr>
<td>B</td>
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<td>88.3</td>
<td>89.3</td>
<td>102.5</td>
</tr>
<tr>
<td>C</td>
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<td>154.8</td>
<td>173.4</td>
<td>187.8</td>
<td>180.9</td>
</tr>
<tr>
<td>D</td>
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<td>145.1</td>
<td>212.3</td>
<td>234.0</td>
</tr>
<tr>
<td>E</td>
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<td>97.5</td>
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<tr>
<td>F</td>
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</tr>
<tr>
<td>G</td>
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<td>126.8</td>
<td>142.3</td>
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<tr>
<td>Mean ± SD</td>
<td>80.56 ± 14.36</td>
<td>125.69 ± 33.93</td>
<td>133.65 ± 40.36</td>
<td>137.90 ± 29.34</td>
<td>157.47 ± 43.57</td>
<td>164.97 ± 43.76</td>
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P value

- N.A.: not applicable.

**DISCUSSION**

In this study, we have demonstrated the feasibility of CS-3D MERGE for black-blood MRI. The total variation (TV) penalty and identity transform based CS algorithm was used. As mentioned in previous studies (9,16), TV is able to effectively remove noise and yield sharp edges in the reconstructed image and thus benefits the recovery of vessel wall for undersampled datasets. In addition, as the vessel wall has relatively high contrast to other parts in black-blood imaging and the proportion of the object to the whole image is very small, vessel wall imaging can be considered sparse in the pixel domain. Therefore, the identity transform was selected as the sparsity transform. According to the CS theory, the relatively high contrast can be recovered with very high probability by $\ell^1$-norm minimization (11). This suggests that the vessel wall signal can be recovered effectively in undersampled k-space. Moreover, the results of in vivo experiments have verified this point, even at high accelerations.

**FIG. 4.** Representative images from one subject. **a:** Three axial slices showing the vessel wall with six levels of acceleration. Almost no difference could be found in blood suppression between the images at acceleration factors of 2, 2.5, 3, 4, and 5 and the corresponding fully sampled image (acceleration 1x). The delineation of vessel wall in the undersampled images is comparable with the corresponding fully sampled image, except for the slightly blurred vessel wall boundary at 5x acceleration. **b:** The reformats of images in oblique planes demonstrated good vessel wall delineation of right (top row) and left (bottom row) carotid arteries. The comparable quality of all the undersampled images can be seen at the carotid bifurcation, where carotid atherosclerosis usually occurs due to plaque deposition.

**FIG. 5.** Representative images from one subject scanned with an isotropic high spatial resolution of $0.7 \times 0.7 \times 0.7$ mm$^3$, and the resolution were zero interpolated to $0.35 \times 0.35 \times 0.35$ mm$^3$. **a:** Three axial slices showing the vessel wall. Comparable flow suppression efficiency and delineation of outer vessel wall boundary and lumen-wall interface are found for each acceleration. **b:** Oblique reformats of images showing comparable vessel wall delineation of the right carotid artery. The carotid bifurcations can be depicted clearly for all accelerations.
Previous studies have reported that centric phase encoding order places the greatest black-blood effect at the center of k-space during acquisition (17,18). In this study, we demonstrated that a pseudo-centric phase encoding order effectively suppresses flowing signals in undersampled 3D k-space. This is vital to black-blood applications because effective flowing blood signal suppression is critical for precise delineation of the lumen-wall interface to achieve high quality imaging of the vessel wall (19–21). Moreover, we found that the flow suppression efficiency for all undersampled images at the carotid bifurcation is comparable with fully sampled images (Figs. 4 and 5), which further demonstrated the capability of the technique.

In the study, we used a slice thickness of 0.7 mm to determine the feasibility of the proposed technique for isotropic high spatial resolution imaging that usually has low signal-to-noise ratio. As comparable image quality was observed between undersampled and fully sampled images, the proposed technique can be used to conduct vessel wall imaging with either low or high spatial resolutions regardless of the different noise levels.

There were several limitations in our study. First, subtle blurring was present in the outer vessel wall and lumen-wall interface when the acceleration was higher than 4-fold. It may suggest that imaging with high acceleration may only be suitable for clinical screening purposes such as carotid atherosclerosis detection. Second, multiple time point scans are required to demonstrate reproducibility of black-blood MRI by the proposed method. Third, as patients with carotid artery stenosis were not studied, it is currently unknown if the undersampled method can provide comparable morphological measurements of plaque to fully sampled datasets. Finally, comparisons between CS and other undersampling techniques such as parallel-imaging, partial Fourier methods have not been implemented in black-blood imaging. These comparisons may be performed in future study to further evaluate which acceleration scheme can provide the best image quality for vessel wall.

CONCLUSION

The CS-3D MERGE technique, which uses undersampled image acquisition, provides flow suppression efficiency and vessel wall image quality which are comparable with fully sampled acquisitions. As this technique offers significant improvement of scan efficiency, it could be a valuable tool for rapid vessel wall imaging.

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REFERENCES