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Autocalibrating Motion-Corrected Wave-Encoding for Highly Accelerated Free-Breathing Abdominal MRI

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Abstract

Purpose—To develop a motion-robust wave-encoding technique for highly accelerated free-breathing abdominal MRI.

Methods—A comprehensive 3D wave-encoding-based method was developed to enable fast free-breathing abdominal imaging: (a) auto-calibration for wave-encoding was designed to avoid extra scan for coil sensitivity measurement; (b) intrinsic butterfly navigators were used to track respiratory motion; (c) variable-density sampling was included to enable compressed sensing; (d) golden-angle radial-Cartesian hybrid view-ordering was incorporated to improve motion robustness; (e) localized rigid motion correction was combined with parallel imaging compressed sensing reconstruction to reconstruct the highly accelerated wave-encoded datasets. The proposed method was tested on six subjects and image quality was compared with standard accelerated Cartesian acquisition both with and without respiratory triggering. Inverse gradient entropy and normalized gradient squared metrics were calculated, testing whether image quality was improved using paired *t*-tests.

Results—For respiratory-triggered scans, wave-encoding significantly reduced residual aliasing and blurring compared with standard Cartesian acquisition (metrics suggesting $P < 0.05$). For non-respiratory-triggered scans, the proposed method yielded significantly better motion correction compared with standard motion-corrected Cartesian acquisition (metrics suggesting $P < 0.01$).

Conclusion—The proposed methods can reduce motion artifacts and improve overall image quality of highly accelerated free-breathing abdominal MRI.

Keywords

wave-encoding; motion correction; abdominal imaging; acceleration; parallel imaging; auto-calibration

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Introduction

Abdominal MRI is often sensitive to various sources of motion, such as respiratory motion, cardiac motion, and subject movements. Breath-holding techniques are commonly used to eliminate respiratory motion when applicable (1–3). However, breath-held scans become impractical for uncooperative patients, such as young children, or when scan times are too long.

To enable free-breathing abdominal MRI, a variety of techniques have been developed. Respiratory gated or triggered data acquisition based on bellows or navigators achieves motion-robust images by acquiring signals from a particular motion state (4). However, this approach usually results in long scan times because of low acquisition efficiency. To enable high acquisition efficiency, non-triggered free-breathing acquisition approaches using Cartesian or non-Cartesian sampling have been developed. Cartesian approaches, such as radial-Cartesian hybrid imaging (5, 6), usually achieve better motion properties through motion-robust view ordering. Non-Cartesian trajectories, such as PROPELLER (7) and radial trajectories (8–13), are also used to improve the motion robustness of abdominal imaging. Motion correction techniques, such as k-space segmentation (14), image-registration (15,16), soft-gating (5,14), and autofocusing (17), have also been combined with these techniques to reduce motion artifacts.

To further shorten the scan time, acceleration methods based on under-sampling and nonlinear reconstruction have been developed in recent studies (18–38). Among these methods, parallel imaging uses phase-array coils with variant sensitivities to recover the missing k-space samples (18–22,24,25,27,31). Compressed sensing exploits the sparsity of MR images to perform constrained reconstruction (23, 26, 30). Combination of these two approaches often yields better reconstruction performance (29,33–39). However, in parallel imaging, the maximum potential acceleration factor is limited by coil geometry, and in practice further reduced by not utilizing coil sensitivity variations in the readout direction (18,19,40).

Wave-encoding has been demonstrated in a recent study to significantly improve the image quality and lower the g-factor in parallel imaging with efficient use of the sensitivity variations in all spatial axes (41). It has also been combined with compressed sensing, and it has been shown to significantly reduce aliasing artifacts in brain imaging (42,43). However, two potential problems with this method may hinder its application to abdominal imaging. First, a separate calibration scan is usually required to estimate coil sensitivity maps (41). Subject motion from breathing between the calibration scan and the wave-encoded acquisition may introduce sensitivity variations and, therefore, may degrade the quality of image reconstruction (19,20). Recently, self-calibrated wave-CAIPI has been introduced (44). However, for random sampling, it is time-consuming to jointly estimate sensitivity maps and the point spread function for wave (wave-PSF). Second, the wave-encoded 3D acquisition may be corrupted by intra-scan subject motion, which limits its performance for free-breathing abdominal imaging.

To solve these problems, we develop a wave-encoding technique for highly accelerated imaging of the abdomen in a 3D Cartesian framework and assess its motion robustness. The main contributions of this work include: (a) self-navigated motion estimates are obtained from the data acquired with wave-encoding using Variable Density sampling and Radial view ordering (VDRad) (5); (b) a novel auto-calibration technique is developed for wave-encoded MRI to improve the coil sensitivity estimation in presence of motion; (c) a localized rigid motion correction method for wave-encoding is derived to further improve motion robustness. The comprehensive imaging technique for abdominal imaging is tested and compared with standard accelerated Cartesian acquisition in in-vivo studies, in which the overall image quality of both methods is assessed.

Theory

Summary of wave-encoding

As illustrated in Fig. 1a, wave gradients are added to 3D Cartesian acquisition during readout to achieve wave-encoding. A previous study has shown that the wave-encoded signal and the underlying magnetization have the following relationship in the k_x - y - z domain (41):

$$S_{\text{wave}}[k_x, k_y, k_z] = \mathcal{F}_y \mathcal{F}_z \{ \text{PSF}[k_x, y, z] \cdot \mathcal{F}_x \{ m[x, y, z] \} \} \quad [1]$$

where $S_{\text{wave}}[k_x, k_y, k_z]$ denotes the k-space data acquired with wave-encoding at location $[k_x, k_y, k_z]$, $\mathcal{F}_x$, $\mathcal{F}_y$, and $\mathcal{F}_z$ denote the DFT operators in x, y, and z axes, $\text{PSF}[k_x, y, z]$ denotes the wave-PSF corresponding to the wave gradients, and $m[x, y, z]$ denotes the underlying magnetization at location $[x, y, z]$. The PSF can be expressed as:

$$\begin{aligned} \text{PSF}[k_x, y, z] &= \exp \left(-i\gamma \int_0^{\tau[k_x]} (g_y(t) \cdot y + g_z(t) \cdot z) dt \right) \\ &= \exp (C_1[k_x] \cdot y + C_2[k_x] \cdot z) \end{aligned} \quad [2]$$

where $\tau[k_x]$ is the time corresponding with the discretized readout point k_x , $g_y(t)$ and $g_z(t)$ are amplitude of wave gradients at time t , $C_1[k_x] = -i\gamma \int_0^{\tau[k_x]} g_y(t) dt$, and $C_2[k_x] = -i\gamma \int_0^{\tau[k_x]} g_z(t) dt$.

As shown previously (42,43), with the estimated coil sensitivity maps and wave-PSF, a modified SENSE reconstruction with sparsity constraints can be used to recover the unacquired points in a Cartesian framework without gridding. The minimization problem can be written as:

$$\underset{m}{\operatorname{argmin}} \left\{ \|D \mathcal{F}_y \mathcal{F}_z P \mathcal{F}_x E m - b\|_2^2 + \lambda R(m) \right\} \quad [3]$$

where D is a k-space sampling operator, P is the wave-PSF, E is the coil sensitivities, m is the desired image, and b is the partially acquired wave-encoded k-space. R is a sparsity transform with regularization coefficients λ .

Auto-calibrated estimation of coil sensitivity maps

Auto-calibration helps reduce the reconstruction errors due to motion-corrupted coil sensitivity estimates (19, 20, 45). Wave-encoded MRI does not directly provide a Cartesian auto-calibration region for parallel imaging. However, by fully sampling the center k-space (Fig. 1b), it is possible to demodulate the wave-encoded center k-space with the known wave-PSF parameters $C_1[k_x]$ and $C_2[k_x]$ into a set of low-resolution images. To achieve this goal, the key step is to get the low-resolution wave-PSF using $C_1[k_x]$ and $C_2[k_x]$, which has the capability to demodulate the calibration region. According to Eq. 2, the PSF associated with a particular spatial resolution δy and δz can be calculated by discretizing the spatial positions y and z with δy and δz , as follows:

$$\begin{aligned} PSF[k_x, y[i], z[j]] &= \exp(C_1[k_x] \cdot y[i] + C_2[k_x] \cdot z[j]) \\ &= \exp(C_1[k_x] \cdot \delta y \cdot i + C_2[k_x] \cdot \delta z \cdot j) \quad [4] \end{aligned}$$

where i and j denote the index of pixels in y and z axes. In this sense, the wave-PSF for any given spatial resolution can be generated with $C_1[k_x]$ and $C_2[k_x]$. In the specific case of the calibration k-space, δy and δz are the resolution of the calibration image in y and z axes, and i and j are indices of points within the calibration region. The calculated low-resolution PSF is subsequently used to demodulate the wave-encoded center k-space into a Cartesian k-space. This demodulated k-space region contains coil sensitivity information, and ESPIRiT (46) can be used to estimate the sensitivity maps. An example of the low-resolution PSF and its corresponding high-resolution PSF are shown in Fig. 1c.

Localized motion correction

Linear phase correction is used to perform localized motion correction by assuming the non-rigid motion of tissue is composed of small regions of linear translations (17). As shown in Eq. 1, the wave-encoded k-space is not a 3D DFT of the image. Therefore, conventional phase correction for linear translation derived from the Fourier transform theory is no longer applicable. Here, a new linear phase correction for wave-encoded k-space is derived.

Assuming $d_x[k_x]$, $d_y[k_x]$, and $d_z[k_x]$ are the estimated distance of linear translation in three orthogonal directions at time $\tau[k_x]$, and the directly acquired image is $m[x, y, z]$, the corrected image will be $m[x - d_x, y - d_y, z - d_z]$. For simplicity, the index k_x is omitted in following derivations. According to Eq. 1, the corrected wave-encoded k-space associated with $m[x - d_x, y - d_y, z - d_z]$ will be

$$\begin{aligned} S_{\text{wave}}^c[k_x, k_y, k_z] &= \mathcal{F}_y \mathcal{F}_z \{ PSF[k_x, y, z] \cdot \mathcal{F}_x \{ m[x - d_x, y - d_y, z - d_z] \} \} \\ &= \mathcal{F}_y \mathcal{F}_z \{ \exp(C_1 y + C_2 z) \cdot \mathcal{F}_x \{ m[x - d_x, y - d_y, z - d_z] \} \} \quad [5] \end{aligned}$$

By reformatting the expression into the wave-PSF corresponding with $m[x - d_x, y - d_y, z - d_z]$, we have

$$\begin{aligned}
S_{\text{wave}}^c[k_x, k_y, k_z] &= \exp(C_1 d_y + C_2 d_z) \cdot \mathcal{F}_y \mathcal{F}_z \{ \exp(C_1 (y - d_y) + C_2 (z - d_z)) \\
&\quad \cdot \mathcal{F}_x \{ m[x - d_x, y - d_y, z - d_z] \} \} \\
&= \exp(C_1 d_y + C_2 d_z) \cdot \mathcal{F}_y \mathcal{F}_z \{ \text{PSF}[k_x, y - d_y, z - d_z] \\
&\quad \cdot \mathcal{F}_x \{ m[x - d_x, y - d_y, z - d_z] \} \}
\end{aligned} \tag{6}$$

According to the spatial-shift theory in DFT, by taking DFT in x, y, z axes respectively, the original wave-encoded k-space can be expressed with respect to $m[x - d_x, y - d_y, z - d_z]$ as

$$\begin{aligned}
S_{\text{wave}}[k_x, k_y, k_z] &= \mathcal{F}_y \mathcal{F}_z \{ \text{PSF}[k_x, y, z] \cdot \mathcal{F}_x \{ m[x, y, z] \} \} \\
&= \mathcal{F}_y \mathcal{F}_z \{ \text{PSF}[k_x, y - d_y, z - d_z] \cdot \mathcal{F}_x \{ m[x - d_x, y - d_y, z - d_z] \} \} \\
&\quad \cdot \exp(i2\pi \cdot (k_x \cdot d_x + k_y \cdot d_y + k_z \cdot d_z))
\end{aligned} \tag{7}$$

By integrating Eq. 6 and Eq. 7, the relation between the corrected k-space and the original k-space becomes

$$\begin{aligned}
S_{\text{wave}}^c[k_x, k_y, k_z] &= S_{\text{wave}}[k_x, k_y, k_z] \cdot \exp(C_1 [k_x] \cdot d_y [k_x] + C_2 [k_x] \cdot d_z [k_x]) \\
&\quad \cdot \exp(-i2\pi \cdot (k_x \cdot d_x [k_x] + k_y \cdot d_y [k_x] + k_z \cdot d_z [k_x]))
\end{aligned} \tag{8}$$

where $S_{\text{wave}}^c[k_x, k_y, k_z]$ is the corrected k-space readout at phase-encoding $[k_y, k_z]$, while $S_{\text{wave}}[k_x, k_y, k_z]$ is the corresponding original k-space readout. This equation shows the wave-encoded k-space can be corrected with linear phase multiplications in phase-encodes. Here, the additional multiplication of $\exp(C_1 d_y + C_2 d_z)$ compensates for phase shifts due to linear translation of the wave-PSF.

Methods

Data acquisition

Calibration of wave-PSF—A calibration scan for estimating the wave-PSF can be performed on a phantom or the scan subject under the same acquisition parameters before or after the routine scan. The calibration scan contains four phases of 2D acquisitions to calibrate each wave gradient respectively. As described in the previous study (41), by phase unwrapping and regression with respect to each position $[y, z]$ for each readout time point, the parameters $C_1[k_x]$ and $C_2[k_x]$ were estimated for each k_x . This PSF can be reused for scans under the same geometry prescriptions regardless of subject motion between scans.

Wave-encoding with VDRad sampling ordering—Wave gradients were applied during the readout with seven cycles and 8.0 mT/m amplitude. A readout oversampling factor of three was used in all scans with spatially selective excitations in the readout direction. The complete sequence diagram is illustrated in Fig. 1a. Other acquisition parameters are shown in Table 1. VDRad sampling ordering (5) was used to achieve a

variable density sampling with a capability to maintain a variable-density pseudorandom subsampling for its subsets. In detail, phase-encodes were grouped into spokes, and spokes were acquired sequentially according to the golden angle (Fig. 1b) (47). A 20×20 region was fully sampled in the center of k-space for auto-calibrated estimation of coil sensitivity maps (Fig. 1b).

Motion estimation with butterfly navigators—For free-breathing continuous acquisition, 3D translation motion was estimated using butterfly navigators (17), which are a form of intrinsic navigators that have minimal time penalty. These navigators were built into the prewinders of each readout, in between the RF pulses and wave-encoding gradients (Fig. 1a). The 3D translation motion estimates are extracted every three TR for every coil element. An example of respiratory motion estimates is shown in Fig. 1c. This enables localized motion correction and soft-gating based on the amounts of motion.

Accelerated reconstruction with soft-gating and localized motion correction

In this work, a combination of parallel imaging and compressed sensing was used to recover the unacquired and corrupted k-space samples. Eq. 3 shows a modified SENSE reconstruction with sparsity constraints for recovering unacquired points. This reconstruction was modified with (a) an additional weighting term to reduce the influence of acquisitions with large amounts of motion, and (b) a localized motion correction of the diaphragm based on the navigator motion measurements. The new reconstruction can be reformatted as an unconstrained optimization problem:

$$\underset{m}{\operatorname{argmin}} \|W (D \mathcal{F}_y \mathcal{F}_z P \mathcal{F}_x E m - \Phi b)\|_2^2 + \lambda \|\Psi m\|_1 \quad [9]$$

where W is a diagonal matrix containing the weights, which acts as a soft-gating (5, 35) of all acquisitions. As illustrated in Fig. 1c, these weights are estimated based on the k-space location and the motion measurements given by the butterfly navigators (5, 35). Φ is the operator for localized motion correction of the acquired k-space b . In detail, for motion estimates $(d_x[k_x], d_y[k_y], d_z[k_z])$, the corresponding k-space signal was multiplied with $\exp(C_1[k_x] \cdot d_y[k_y] + C_2[k_x] \cdot d_z[k_z]) \cdot \exp(-i2\pi \cdot (k_x \cdot d_x[k_x] + k_y \cdot d_y[k_y] + k_z \cdot d_z[k_z]))$, according to Eq. 8. The motion of the diaphragm was estimated using the anterior coil element near the diaphragm. Ψ is a wavelet transform operator and the term $\|\Psi m\|_1$ promotes the sparsity of the solution. λ is the regularization parameter related to the ℓ_1 -norm of the wavelet coefficients of m . This parameter was respectively tuned and set empirically to be 0.005 for wave-encoding and 0.01 for Cartesian. Fast iterative shrinkage-thresholding algorithm (FISTA) (48) was used to solve the optimization problem in Eq. 9.

Simulation and phantom study

To find the optimal calibration size for auto-calibrated estimation of sensitivity maps and estimate the relative signal-to-noise ratio (SNR), fully-sampled wave-encoded k-space and fully-sampled Cartesian k-space were acquired with a quality-assurance phantom on a GE MR750 3T scanner (GE Healthcare, Waukesha, WI) using an 8-channel cardiac coil and a 3D spoiled gradient echo (SPGR) acquisition sequence. The acquisition parameters for this

sequence are shown in Table 1. The sensitivity maps for wave-encoding were estimated with different sizes of wave-encoded calibration k-space (8×8 to 96×96) using the proposed auto-calibration method. Since the phantom remains static during the entire scan, the sensitivity maps estimated from the Cartesian k-space were used as the reference. The normalized root-mean-squared error (nRMSE) between the sensitivity maps for wave-encoding and Cartesian acquisition was used as a metric for determining the optimal calibration size. Another set of simulated wave-encoded k-space generated from the Cartesian k-space was also tested in the same manner to exclude the influence of wave-PSF estimation error. The SNR of fully-sampled images was estimated using the pseudo random replicas approach (40).

In vivo study

In vivo scan setup—With Institutional Review Board approval and informed consent, in vivo studies were performed on six healthy adults (five male, one female, ranging from 22 to 24 years). Studies were performed on a GE MR750 3T scanner using a 32-channel torso coil and four different 3D SPGR acquisition sequences. These sequences were acquired for each subject in random orders, including sequences with/without wave-encoding and with/without respiratory triggering. Subjects were not informed of the sequence type or ordering. The acquisition parameters for these sequences are shown in Table 1. For better performance, longer TE/TR were used in wave-encoded acquisition. Optimal flip-angles for all sequences was determined according to reference (49). Fat-suppression was incorporated with a periodic spectral fat-inversion pulse with an inversion time of 9.0 ms and approximately 25 views following each fat-inversion pulse. To keep the same workflow, scan time for all these sequences was set to be equal by adjusting the acceleration factor for each sequence. For respiratory-triggered acquisitions, an external respiratory bellow was used to trigger the acquisition at the end of exhalations, and no soft-gating or motion correction were used in the corresponding reconstruction. For free-breathing continuous acquisition, motion was estimated from the navigators and reconstruction was conducted with soft-gating and localized motion correction for both Cartesian and wave-encoded k-space. All the images were reconstructed using a combination of C/C++ (the BART toolbox (50)) and MATLAB (MathWorks, Natick, MA).

Evaluation of image quality—As suggested in the comparison of metrics for motion-corrupted image quality (51), two metrics with good capability to quantify motion-corrupted image quality were implemented in this work to evaluate the image quality of different

approaches. They are inverse gradient entropy (IGE) given by $-\left(\sum_{i,j} h_{i,j} \log_2 [h_{i,j}]\right)^{-1}$,

where $h_{i,j} = |g_{i,j}| / \sum_{i,j} |g_{i,j}|$, and normalized gradient squared (NGS) given by

$\sum_{i,j} \left(|g_{i,j}| / \sum_{i,j} |g_{i,j}| \right)^2$, where $g_{i,j}$ is gradient at coordinate i,j . Higher value of these metrics indicate better image quality and fewer motion artifacts. Before calculating these metrics, the images were normalized. The metrics were calculated for a specific region of interest (ROI) of approximately 120×140×100 mm³ containing the liver. The calculated value of these metrics were also normalized with respect to the free-breathing non-motion-

corrected results. Respiratory-triggered and free-breathing results were evaluated using paired *t*-tests respectively. P-value of under 0.05 was considered as statistically significance.

Results

Simulation and phantom study

An analysis of the accuracy of auto-calibrated estimation of coil sensitivities (Fig. 2a) was performed in the ideal case with simulated wave-encoded k-space. The normalized RMSE decreased to under 0.015 when the calibration size was no less than 20×20 . When the calibration size was larger than 40×40 , the normalized RMSE went below 0.005 and reached a plateau. With estimated wave-PSF, due to the error induced by wave-PSF imprecision, the normalized RMSE was approximately 0.001 to 0.002 higher than simulation. The SNR of the fully-sampled wave-encoded acquisition was approximately 1.45 times higher on average than the fully-sampled Cartesian acquisition (Fig. 2b).

In vivo study

Examples of free-breathing respiratory-triggered and free-breathing non-triggered results are shown in Fig. 3 and Fig. 4, respectively. In Fig. 3, both standard Cartesian and wave-encoded acquisition yielded images without noticeable motion artifacts. Compared with Cartesian results, wave-encoding improved the image quality by reducing residual aliasing artifacts (white dotted arrows) and improving the delineation of detailed structures (white solid arrows). The liver appeared to be more uniform and less noisy compared with standard approaches.

Similar performance can be observed in Fig. 4. Images without motion correction were mostly corrupted with blurring. With localized motion correction, motion artifacts were significantly reduced in both Cartesian and Wave-encoded acquisitions. However, in the Cartesian results, residual aliasing artifacts due to soft-gating and residual motion artifacts were still visible compared with the wave-encoding approach.

In Fig. 5a, the image quality for respiratory-triggered scans was significantly improved using wave-encoding, according to the result of a paired *t*-test with $P < 0.05$. The standard respiratory-triggered scan also improved upon the reference scans without motion correction. In Fig. 5b, the paired *t*-test shows using the proposed motion-robust wave-encoding, there was significant improvement in the image quality of the liver with $P < 0.01$. In this case, the improvement of motion-robust wave-encoding upon motion-corrected Cartesian results was more marked than the improvement of motion correction upon non-corrected results, as residual aliasing artifacts degraded the quality of motion-corrected Cartesian scans.

Discussion

Wave-encoding enables parallel imaging to exploit coil sensitivity variations in all dimensions (41). Although k-space samplings are acquired in non-Cartesian trajectories, the reconstruction can be performed efficiently in a Cartesian framework. When combined with compressed sensing, it shows higher capability of reducing aliasing artifacts compared with

standard Cartesian reconstruction (42, 43). As a result, in respiratory-triggered scans, the wave-encoding technique significantly improved the quality of reconstructed images. For free-breathing non-triggered scans, the proposed approach uses (a) built-in butterfly navigators to monitor non-rigid motion, (b) auto-calibration to estimate coil sensitivity maps, and (c) localized motion correction to reduce motion artifacts and improve the motion robustness of wave-encoding. For the same scan time, the image quality of free-breathing scans was significantly improved compared with standard Cartesian scans in VDRad ordering.

The amplitudes and the number of cycles of wave gradients were chosen according to reference (41). Due to slew rates and readout bandwidth limits, the readout duration was lengthened to achieve seven cycles of wave-encoding with 8.0 mT/m amplitude. As minimum possible TE and TR were used in both sequences, the TE and TR of wave-encoding were longer than those of standard Cartesian sequence (Table. 1). Therefore, to keep the same scan time, the acceleration factor of the proposed method was set approximately 1.4 times higher than the Cartesian approach. However, in the Cartesian approach, the coil sensitivity variations in the readout axis (superior-inferior here) are not utilized in parallel imaging reconstruction. The 32-channel coils for torso imaging usually have four rows of channels in the readout axis for coronal scans, which results in improvement in image quality after wave-decoding reconstruction despite increasing the acceleration factor by 1.4 times. The increase of TE, TR, flip-angle, and readout duration for wave-encoding also contribute to improvements in SNR in the raw images.

Since the SNR of wave-encoded acquisition may differ from Cartesian acquisition, the regularization coefficients were respectively optimized for wave-encoding and Cartesian when the reconstruction was performed. The contrast of images using wave-encoding and Cartesian acquisition may also be different because of the inconsistency of acquisition parameters. This should be evaluated and optimized in future studies with clinical scans, particularly with intravenous contrast enhancement.

The use of auto-calibration enables estimation of coil sensitivities from acquired 3D wave-encoded k-space. Note that the size of calibration region may affect the accuracy of estimated coil sensitivity maps. Larger calibration size results in higher accuracy of sensitivity estimation, but also leads to longer acquisition and lower acceleration factor. In our experiments, the calibration size is chosen to be approximately 20×20 for both Cartesian and wave-encoded k-space to achieve a trade-off between coil sensitivity accuracy and scan time.

The use of soft-gating and localized motion correction reduces the motion artifacts in reconstructed images. By using soft-gating, phase-encodes are weighted with respect to their amounts of motion. This helps reduce the artifacts from mis-corrected k-space data under the localized motion correction assumptions. Through localized motion correction, motion artifacts are reduced in the ROI of the diaphragm and the liver. The motion estimates used to correct the motion-corrupted k-space can be selected automatically. Based on our experience, finding the biggest distance of superior-inferior motion among all the motion estimates given by butterfly navigators yields motion estimates that can correct the motion in

the diaphragm. Furthermore, repeating localized motion correction using motion estimates from each individual channel can provide globally motion-corrected images by performing localized autofocusing (17).

The use of wave-gradients creates voxel-spreading effects (41). This reduces the noticeable streaking artifacts in the reconstructed images when the combined parallel imaging and compressed sensing reconstruction cannot fully remove the aliasing. Nonetheless, an inaccurate wave-PSF may degrade the performance of wave-encoded reconstruction as well as localized motion correction, as the error in wave-PSF may cause blurring and ghosting artifacts in the final images. Therefore, it is important to get an accurate wave-PSF before performing reconstruction and localized motion correction.

There are several limitations of the proposed method for clinical application. First of all, unlike the auto-calibrated estimation of coil sensitivities, the wave-PSF requires an additional calibration scan, which may take 10-30 seconds, determined by the accuracy of PSF needed. This needs to be repeated in order to be applied to subjects with different geometry prescriptions. The joint k-space trajectory and parallel imaging optimization method (44) provides an insight of calibrationless wave-encoded reconstruction for uniform sampling patterns. Further extension of this method may be helpful for variable-density sampling patterns. Secondly, geometric-decomposition coil compression (52) cannot be used in wave-encoded reconstruction. Therefore, the reconstruction time is approximately 10 times longer (300s on average for a $300 \times 300 \times 100$ acquisition matrix with 32 channels reconstructed on a 48-core 2.3GHz Intel Xeon E5-2670 PC with 256GB RAM and a GeForce GTX TITAN X GPU) than for Cartesian (30s on average for the same acquisition matrix size and hardware). Lastly, the auto-calibration estimation of coil sensitivities may induce additional error in the coil sensitivities when compared to estimation based on Cartesian k-spaces. These limitations motivate the need to develop a wave-PSF calibration-free motion-robust wave-encoding method in the future.

In conclusion, the proposed motion-robust wave-encoding method can improve fast free-breathing abdominal MRI in both respiratory-triggered and non-triggered cases with significantly better image quality.

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Figure 1.

Accelerated free-breathing abdominal MRI method using motion-robust wave-encoding. **a:** Pulse sequence diagram for wave-encoding with an example of butterfly navigator in x axis. In this case, the G_x prewinder is ahead of G_y and G_z prewinders to encode motion in the phase of acquired signals. For butterfly navigators in y or z axis, G_y or G_z prewinder is ahead of the other prewinders (17). Spatially selective excitation is applied in the readout direction. **b:** Illustration of the sampling pattern for auto-calibrated wave-encoding with VDRad (5). The center calibration region is used to estimate the coil sensitivity maps. **c:** Overview of the framework of the proposed method. This includes motion estimation, correction, and soft-gated image reconstruction. The top-left plots show motion estimates for one coil and the corresponding weights used in soft-gating.

Figure 2.

a: Analysis of the effect of calibration size on the normalized RMSE of auto-calibrated sensitivity estimation; the RMSE of the simulation study with an accurate wave-PSF and the RMSE of the phantom study with estimated wave-PSF are shown with blue and red curves, respectively. Normalized RMSE of under 0.015 was achieved when the calibration size was no less than 20×20 . **b:** The SNR ratio between Cartesian and wave-encoded phantom scans ($\text{SNR}_{\text{wave}}/\text{SNR}_{\text{Cartesian}}$). The average SNR ratio inside the phantom was 1.45.

Figure 3.

Example images of free-breathing respiratory-triggered scans using standard Cartesian sampling (left) and wave-encoding (right) on **a:** a 23-year-old male volunteer, **b:** another 23-year-old male volunteer, **c:** a third 23-year-old male volunteer, and **d:** a 22-year-old male volunteer. Reduced aliasing artifacts (white dotted arrows) and improved structure delineation (white solid arrows showing blood vessels) were achieved using the proposed wave-encoding approach.

Figure 4.

Example images of free-breathing non-triggered scans using standard Cartesian sampling (left and middle) without/with motion-correction (MC) and wave-encoding (right) with MC on **a**: a 23-year-old female volunteer, **b**: another 23-year-old male volunteer, **c**: a third 23-year-old male volunteer, and **d**: a fourth 23-year-old male volunteer. Residual aliasing artifacts (white dotted arrows) and residual motion artifacts (white solid arrows) were still visible in Cartesian results, while the proposed motion-corrected wave-encoding approach reduces these artifacts.

Figure 5.

Comparison of the inverse gradient entropy and the normalized gradient squared for the standard Cartesian acquisition and proposed method in **a**: free-breathing respiratory-triggered (RT) scans and **b**: in free-breathing non-triggered (FB) scans of six subjects for an ROI of approximately $120 \times 140 \times 100 \text{ mm}^3$ containing the liver. Statistically significant different results with $P < 0.05$ and $P < 0.01$ are marked with (*) and (**) respectively.

Table 1

Scan Parameter Summary

	Phantom study for auto-calibration evaluation		Phantom study for SNR estimation		In vivo study for adult volunteers	
	Standard Cartesian	Proposed	Standard Cartesian	Proposed	Standard Cartesian	Proposed
TE/TR	1.9/5.4 ms		1.5/3.6 ms	1.9/5.4 ms	1.5/3.6 ms	1.9/5.4 ms
#TR	16384		23232		6200±50	4400±50
Flip Angle	18°		15°	18°	15°	18°
Readout bandwidth	±111.1 Hz/pixel		±333.3 Hz/pixel	±111.1 Hz/pixel	±333.3 Hz/pixel	±111.1 Hz/pixel
Resolution	1.0×1.9×2.0 mm ³		1.1×1.1×2.4 mm ³		(1.1±0.1)×(1.1±0.1)×(2.3±0.1) mm ³	
FOV	30.0×24.0×25.6 cm ³		34.0×34.0×24.0 cm ³		(32.0±2.0)×(32.0±2.0)×(23.0±1.0) cm ³	
Coil	8-channel Cardiac		8-channel Cardiac		32-channel Torso	
Reduction factor	Fully-sampled		Fully-sampled		2.5 × 2.0	3.2 × 2.2
Scan time	58 s		104 s	154 s	110±5 s (Respiratory-triggering) 34±2 s (Free-breathing)	
Maximum S/I motion	0		0		15±5 mm	
Fat suppression	Spectral fat-inversion with inversion time of 9ms					