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Magnetization-Prepared Shells Trajectory with Automated Gradient Waveform Design

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Abstract

PURPOSE—To develop a fully-automated trajectory and gradient waveform design for the non-Cartesian Shells acquisition, and to develop a magnetization-prepared Shells acquisition to achieve an efficient three-dimensional (3D) acquisition with improved gray-to-white brain matter contrast.

METHODS—After reviewing the Shells k-space trajectory, a novel, fully-automated trajectory design is developed that allows for gradient waveforms to be automatically generated for specified acquisition parameters. Designs for two types of Shells are introduced, including fully-sampled and undersampled/accelerated Shells. Using those designs, a magnetization-prepared Shells (MP-Shells) acquisition is developed by adjusting the acquisition order of shells interleaves to synchronize the center of k-space sampling with the peak of desired gray-to-white matter contrast. The feasibility of the proposed design and MP-Shells is demonstrated using simulation, phantom, and volunteer subject experiments, and the performance of MP-Shells is compared with a clinical Cartesian magnetization-prepared rapid gradient echo acquisition (MP-RAGE).

RESULTS—Initial experiments show that MP-Shells produces excellent image quality with higher data acquisition efficiency and improved gray-to-white matter contrast-to-noise ratio (by 36%) compared with the conventional Cartesian MP-RAGE.

CONCLUSION—We demonstrated the feasibility of a 3D MP-Shells acquisition and an automated trajectory design to achieve an efficient acquisition with improved gray-to-white matter contrast.

Keywords

non-Cartesian MRI; Shells trajectory; magnetization prepared; parallel imaging

INTRODUCTION

While most routine clinical MR imaging is based on Cartesian sampling, the benefits of various non-Cartesian sampling strategies have been demonstrated in a variety of

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applications (1–8). Shells trajectory is a three-dimensional (3D) non-Cartesian acquisition method that samples k-space using a series of concentric spherical shells (2,9–11). A collection of interleaved helical spiral readouts traverse the surfaces of k-space shells from one pole to the other, which together enable k-space to be completely sampled. The unique geometry of the Shells trajectory yields several inherent advantages. First, Shells naturally performs k-space corner cutting in all three dimensions – an efficient acceleration strategy that does not substantially compromise image quality (9). This strategy yields an approximately 2× scan time reduction compared to a fully sampled conventional 3D Cartesian scan covering a cubic volume of k-space centered around the origin, and an approximately 1.5× scan time reduction compared to the 3D Cartesian acquisition employing corner cutting within its 2D plane formed by the two phase encoded directions. Second, since the number of spiral interleaves can be independently varied on a per-shell basis, the sampling density can be changed as a defined function of the radii of shells to achieve further acceleration. The Shells trajectory was shown to be well-suited for undersampling as it possesses non-coherent aliasing artifacts (9). Third, the spherical k-space volume surrounding the center of k-space can be sampled with only several readouts. As the k-space center contains the low spatial frequency components, which primarily determine the image signal level and contrast properties, the Shells trajectory is ideal for capturing the transient evolution of image contrast in a 3D acquisition with optimal efficiency (12). Furthermore, the ordering of the shells sampling as a function of the radius can be flexibly arranged to adapt to specific applications. For contrast-enhanced MR angiography, the acquisition of the central shells in k-space can be synchronized to the arrival of the first pass of a contrast agent at the artery of interest, suppressing venous phase signal, as demonstrated previously in the intracranial venous suppressed contrast-enhanced MR angiography (10).

Magnetization preparation is another class of clinical MR applications that requires high k-space center sampling efficiency to capture the transient contrast. The magnetization prepared rapid gradient echo sequence (MP-RAGE) (13) has been widely used for high-resolution whole brain T1-weighted imaging, such as brain morphometry studies for a series of neurodegenerative diseases (14–17), study of early brain development and pathology (18,19), and evaluation of therapy or treatment response (20,21). In MP-RAGE, a 180-degree magnetization preparation (inversion) RF pulse is applied before each train of data acquisitions to increase the contrast between two tissue types of interest, such as gray and white matters in the brain. The standard 3D Cartesian MP-RAGE brain sequence uses a sequential view order that acquires the central k-space lines so that strong gray-to-white contrast is expected (13). By adjusting the acquisition order of the shells and their interleaves, i.e. synchronizing the center of shells sampling with the peak of the desired contrast, we hypothesize that the Shells can be specifically designed to handle the evolution of gray-to-white matter contrast in a time-efficient manner (22). In this work, this special case of Shells design, which is termed MP-Shells, is presented.

The Shells trajectory has historically been implemented using a heuristic gradient waveform design strategy that required some manual intervention (9,10). When isotropically-sampled 3D spiral interleaves were used to sweep across a shell's surface from one pole to the other, it was previously necessary to stop the trajectories when they approached the so-called

“polar icecap” regions (i.e., the circular, dome-shaped areas around the poles), due to system hardware constraints on the gradient slew rate. To ensure globally sufficient k-space sampling, two separately-designed, helical spiral patches were additionally used to cover these polar icecap regions. This disjoint sampling approach is not only inefficient but can cause phase discontinuities between data acquired in different interleaves (10). A later improvement (23) to the Shells trajectory design replaced the polar icecap acquisitions with continuous helical spiral interleaves that sample the surface of the nominal shells from pole-to-pole. However, unlike the primary shells, these extra interleaves required a three-stage manual design for the spiral, radial, and connecting trajectory components (9,10). Here, building on preliminary work (22,24), we demonstrate a completely automated and flexible Shells trajectory design strategy that allows for different selected imaging settings including field-of-view (FOV), receive bandwidth (BW), spatial resolution, and gradient system hardware limits. Using this new trajectory design, we present a 3D magnetization prepared (MP) Shells acquisition strategy that optimizes the k-space sampling ordering according to the prescribed inversion time, and improves gray-to-white matter contrast over the conventional gradient echo sequence. The feasibility of the proposed Shells trajectory design and MP-Shells is demonstrated using both phantom and human volunteer subject experiments. Its performance is compared with that of a Cartesian MP-RAGE (13) sequence used routinely in clinical practice. As parallel imaging has been investigated for accelerating MP-RAGE quantitative neuroimaging research protocols (25), the feasibility of the proposed Shells acquisition combined with sensitivity encoding (SENSE)-type parallel imaging reconstruction (26) is also demonstrated in this paper using a series of phantom and *in vivo* experiments.

THEORY

Shells Trajectory Design

Following reference (24), 3D k-space can be sampled using a series of concentric shells with radii of $k_{r,n} = n\Delta k_r$, where $1 \leq n \leq N$ is the index of each k-space shell, N denotes the total number of shells, and Δk_r denotes the radial distance among consecutive shells. For a user-defined imaging FOV and isotropic imaging resolution (Δx), Δk_r and N can be determined

based on the Nyquist sampling requirement as $\Delta k_r = \frac{1}{\text{FOV}}$ and $N = \frac{\text{FOV}}{2\Delta x}$. The n -th k-space shell with a radius of $k_{r,n}$ can be sampled with a series of 3D spiral interleaves expressed using the following parametric function of the time parameter, τ .

$$\begin{aligned} k_{n,m,x}(\tau) &= k_{r,n} \sin\left(\frac{\pi\tau}{T}\right) \cos\left(\frac{\omega_n\tau}{T} + \theta_{n,m}\right) \\ k_{n,m,y}(\tau) &= k_{r,n} \sin\left(\frac{\pi\tau}{T}\right) \sin\left(\frac{\omega_n\tau}{T} + \theta_{n,m}\right) \\ k_{n,m,z}(\tau) &= k_{r,n} \cos\left(\frac{\pi\tau}{T}\right) \end{aligned} \quad (1)$$

where T is a user pre-determined parameter denoting the total number of readout points in each interleave; $\tau (\tau \in [0, T])$ is the input parameter of the curve denoted by Eq. 1, which is discretized uniformly between 0 and T into a total of T samples, with $\tau=0$ corresponds to the initial point of the interleave, and $\tau=T$ corresponds to the end. The parameter ω_n

controls the azimuthal rotational speed of the interleaves/readouts on the n -th shell.

$\theta_{n,m} = \frac{2m\pi}{M_n}$ denotes the starting phase of the m -th interleave ($0 \leq m \leq M_n - 1$) in the azimuthal direction and M_n indicates the total number of interleaves required for the n -th shell.

Assuming rotational invariance, the m -th interleave of the n -th shell can be obtained by azimuthally rotating the gradient waveforms of an initial interleave with zero phase (i.e., $m = 0$) by an angle $\theta_{n,m}$. We therefore first focus on the design of the initial interleave with $\theta_{n,m=0} = 0^\circ$, which is independent from M_n . For a given k-space curve with the parametric equation defined by Eq. 1, the physical x , y , and z gradient waveforms can be automatically calculated using the design method proposed by Lustig et al. (27). This algorithm generates gradient waveforms sampling along a parametrized trajectory in a time-optimal fashion, while satisfying pre-defined, maximum gradient amplitude and slew rate limits. The primary goal of the algorithm is to minimize the time required to sample along a certain trajectory. Consequently, for a certain ω_n in Eq. 1, the actual time of the readout gradient waveforms generated using this method, or equivalently, the number of samples (T_0) for a given bandwidth, may differ from the original designated value, T . However, it is usually desirable to fix the readout time for different interleaves. We therefore iteratively determine the appropriate ω_n for the n -th shell's interleave so that a readout length of the target T is produced. The gradient waveforms of trajectory specified by Eq. 1 are first generated with ω_n initialized as $\omega_{n,i=0} = 2\pi^2 n$ (see the next paragraph for explanation on this choice). The number of readout points, T_i , of the generated gradient waveforms is used to update ω_n

based on $\omega_{n,i+1} = \frac{T}{T_i} \omega_{n,i}$, and gradient waveforms are then regenerated. This process is repeated until a readout length of the target T is obtained. To also ensure that Nyquist sampling criteria are met for each interleave, the input gradient amplitude and slew rate limits are set as the minimum between the gradient hardware constraints and the Nyquist

requirement ($G_{\max} = \frac{2\pi\text{FOV}}{\gamma\text{BW}}$ for each axis).

For an inner shell with a small radius, a single interleave is typically sufficient to sample the shell surface in multiple turns, as illustrated in Fig. 1a. As ω_n determines azimuthal rotation speed which controls the number of turns for the helical spiral, in this case the number of

turns ($N = 2\frac{\omega_n}{2\pi} = \frac{\omega_n}{\pi}$) should be larger than the minimum ($N_{\text{Nyquist}} = \frac{2\pi k_r(n)}{\Delta k_r} = 2\pi n$) required to universally satisfy Nyquist sampling requirements (i.e., $\omega_n \geq 2\pi^2 n$). For outer shells, however, multiple interleaves are usually required (Fig. 1b). The number of interleaves (M_n) required for the n -th shell can be determined after ω_n is obtained by limiting the maximum distance between the two neighboring interleaves to satisfy the Nyquist sampling distance (Δk_r), as detailed in Appendix, which yields:

$$M_n = \begin{cases} 1, & \text{if } \omega_n \geq 2\pi^2 n \\ \lceil 2\pi n \sin\left(\arctan\left(\frac{\pi}{\omega_n}\right)\right) \rceil, & \text{if } \omega_n < 2\pi^2 n \end{cases} \quad (2)$$

where the ceiling operator $\lceil x \rceil$ calculates the smallest integer equal to or greater than x . A flow chart demonstrating the key design steps using the proposed automated Shells trajectory design strategy is summarized in Fig. 2.

MP-Shells

Due to its flexible centric ordering, magnetization preparation can be readily added into the Shells pulse sequence. Similar to a conventional Cartesian MP-RAGE acquisition, during each repetition time (TR) of an MP-Shells acquisition, a set of shells readouts is performed following a 180-degree inversion preparatory RF pulse, as shown in Fig. 3a. Each acquisition segment is followed by a recovery delay period before the next TR starts. As for a conventional MP-type sequence, the TI, TR, acquisition time window and recovery delay period are user-selected parameters based on the desired contrast. To help ensure a “fair” comparison, here we used the same parameters for the MP-Shells acquisition as those used in our clinical MP-RAGE sequence.

In non-Cartesian acquisition such as Shells, the presence of lipids can cause considerable blurring in the reconstructed images if not properly taken into account. To minimize this artifact, spectrally selective fat saturation modules (28) can be sporadically embedded between shells acquisition readouts. These fat saturation modules also provide extra relaxation time for the target water magnetization and therefore can increase image signal amplitude (and thus SNR). As each individual shell readout is relatively short, each fat saturation module can successfully suppress the lipid signal for several (e.g., $n = 2$ to 4) subsequent shells readouts. Therefore, fat saturation can be performed before every n shells readouts to improve data acquisition efficiency. As a preliminary test, we compared the fat saturation performance by changing the number of shells readouts between two subsequent fat saturation modules (i.e., various choices of n), and a range of $n = 2$ to 4 was found to yield good performance with reasonable scan time increase (See Supporting Figure S2). An example of fat saturation module performed before every $n = 3$ shells readouts is described in Fig. 3b.

With a predefined n , time intervals of Shells readout module (T_{Shells}) and fat saturation module (T_{FatSat}), the number of Shells readouts that fits into the prescribed acquisition time window (T_{Acq}) can be calculated as:

$$N_{\text{Shells}} = \left\lfloor \frac{nT_{\text{Acq}}}{nT_{\text{Shells}} + T_{\text{FatSat}}} \right\rfloor \quad (3)$$

The number of TR required to perform the entire acquisition can then be calculated as:

$$N_{\text{TR}} = \left\lceil \frac{N_0}{N_{\text{Shells}}} \right\rceil \quad (4)$$

where N_0 denotes the total number of Shells readouts.

In MP-Shells, the acquisition order of the other shells is arranged so that the inner shells are preferentially acquired closer to the TI to maximize the desired contrast. Based on Eqs. 3 and 4, there are a total of N_{TR} TRs and each TR has N_{Shells} time slots available for interleave acquisition. To determine the interleave acquisition order, all the shells interleaves are initially sorted according to their radii. The interleaves on the same shell are numerically sorted based on the azimuthal rotation angle. After sorting, the first interleave, which belongs to the smallest shell (closest to k-space center) is placed at the TI time for the first TR. The next interleave in the queue is placed at the TI time of the second TR. This continues until the TI acquisition slots for all the TRs are filled. The next available slots to fill would be those closest to TI in all the TRs. This process is repeated until all the slots are filled, with the last N_{TR} shells interleaves filling the very last slots in all the TRs. As a result of this ordering, in each TR, data acquisition starts from the outermost shells at the beginning of the data acquisition window, gradually transits to the center of k-space when approaching TI, and then progresses back to outer k-space at the end of the acquisition window. This is also indicated by the color bar in Fig. 3a, which uses brighter shades to represent interleaves with smaller radii and darker shades to represent interleaves with larger radii. Note that this strategy is conceptually similar to spiral-in/spiral-out setups commonly used in non-Cartesian functional MRI (fMRI) acquisitions (29).

Reconstruction of Fully Sampled Shells

The images acquired with the fully sampled Shells trajectory designed using Eqs. 1 and 2 can be reconstructed using non-iterative gridding-type reconstruction (30). Denoting $\mathbf{A}$ as the (discretized) forward encoding operator assuming fully sampled Shells trajectory, the underlying image vector, $\mathbf{x}$, can be reconstructed from the acquired k-space signal vector, $\mathbf{g}$, using:

$$\mathbf{x} = \mathbf{A}^* \mathbf{D} \mathbf{g} \quad (5)$$

where $\mathbf{D}$ is the diagonal matrix compensating for variable acquisition density, and can be obtained using the method described by Pipe et al. (31); $\mathbf{A}^*$ denotes the adjoint operator of $\mathbf{A}$ and can be implemented using the type-II non-uniform fast Fourier transform (NUFFT) (32,33). Alternatively, the forward operator can take into account the effect of spatial gradient nonlinearity, in which case $\mathbf{A}$ and $\mathbf{A}^*$ can be implemented using a type-III NUFFT (34). For both implementations, the effect of off-resonance can also be corrected using the time-segmentation method (35).

Parallel Imaging using Undersampled Shells Trajectory

Like other non-Cartesian acquisitions, the proposed Shells acquisition can also be combined with parallel imaging to further accelerate data acquisition relative to the intrinsic acceleration of the Shells acquisition. As the outer shells corresponding to the high-frequency components of an image yield k-space signal with less intensity, they can be preferentially undersampled to a higher degree, similar to a variable density acquisition (36–38). The number of interleaves for the n -th shell, Π_n , can be modified by applying a mapping function ($w^{(n)}$) to M_n calculated from Eq. 2, which yields:

$$\Pi_n = w(n) \quad (6)$$

A simple mapping function $w(\cdot)$ can be chosen as:

$$w(n) = \begin{cases} M_n, & \text{for } n \leq n_0 \\ M_{n_0}, & \text{otherwise} \end{cases} \quad (7)$$

so that the number of interleaves reaches a plateau after a pre-defined shell radius (n_0). Like many contemporary accelerated Cartesian acquisition strategies (e.g., DISCO (39)), this undersampling strategy corresponds to a fully-sampled central k-space region with a sampling rate that progressively decreases as a function of radial distance outside of this region.

Denoting $\mathbf{A}_{PI}$ as the forward encoding operator (N_k by N_p) using the undersampled Shells trajectory (Eqs. 1 and 4), the underlying image vector, $\mathbf{x}$, can be reconstructed using the SENSE-type reconstruction framework with Tikhonov regularization by solving the following quadratic optimization problem (26):

$$\arg\min_{\mathbf{x}} \|\mathbf{I}_0 \otimes \mathbf{A}_{PI}\mathbf{S}\mathbf{x} - \mathbf{g}\|_{\mathbf{D}}^2 + \lambda \|\mathbf{x}\|_2^2 \quad (8)$$

where $\mathbf{S}$ represents the sensitivity maps ($N_c \times N_p$ by N_p matrix) of a multi-channel RF receive array, $\mathbf{g}$ is the multi-channel k-space data matrix ($N_c \times N_k$ by 1); N_p , N_c , and N_k are the number of image pixels, the number of receive channels, and the total number of samples for each receive channel, respectively; $\mathbf{I}_0$ is a N_c by N_c identity matrix, $\otimes$ represents the Kronecker product of matrices, λ is the regularization parameter, and $\|\cdot\|_{\mathbf{D}}^2$ denotes the density compensation function ($\mathbf{D}$) weighted norm. Eq. 6 can be solved iteratively using the conjugate gradient (CG) method, where the diagonal matrix $\mathbf{D}$ is commonly interpreted as a preconditioner (31) whose presence improves convergence rates and consequently reduces the number of iterations needed to obtain an acceptable reconstruction result.

METHODS

Phantom Experiment

The American College of Radiology (ACR) phantom was scanned on a 3.0 T GE Signa HDxt system (v16.0) with an 8-channel receive-only brain coil using the proposed fully sampled The Shells trajectory was designed assuming the data acquisition parameters of a typical brain scan volume, with isotropic imaging resolution of $\Delta x = 1.0 \text{ mm}^3$, a spherical FOV with 24 cm diameter, readout length $T = 512$, and readout BW = $\pm 125 \text{ kHz}$. Details of the data acquisition parameters are shown in Table 1. The system gradient amplitude and slew rate limits available on the zoom mode gradient of a 3.0 T GE Signa HDxt system were assumed ($G_0 = 40 \text{ mT/m}$ and $S_0 = 150 \text{ T/m/s}$). The shells readouts were played sequentially

following a center-out order (i.e., inner shells followed by outer shells) in a fashion similar to a small flip angle, spoiled gradient echo (SPGR) acquisition. A 3D dual-echo gradient echo (GRE) B_0 mapping sequence was also performed on the same imaging volume using the following parameters: FOV = 24 cm³, matrix size = 256×128×64, BW = ±31.25 kHz, echo spacing = 2.3 msec, flip angle (FA) = 10°, TR = 9 ms. The point spread function (PSF) of the designed Shells trajectory was first examined. The uniform k-space (i.e., with 3D k-space input $u(k) = 1$ denoting a constant function) was sampled using the designed Shells trajectory, and the obtained k-space signals were reconstructed using the adjoint type-II NUFFT operator (Eq. 5) to calculate the PSF, which confirms the absence of noticeable aliasing artifact within the imaging FOV.

To test the undersampled Shells, the same ACR phantom was also scanned using Shells with different acceleration factors on the same MR system. The numbers of readouts for each shell were determined by applying the mapping function described in Eq. 5 to the number of interleaves required for the fully sampled trajectory design, i.e., Eq. 2. The shell radius thresholds ($n_0 = 38, 47, 61$ in Eq. 5) were chosen to yield an undersampling factor of 2, 3, and 4, respectively.

In Vivo MP-Shells

Under an Institutional Review Board (IRB)-approved protocol, the brain of a healthy volunteer was scanned using the proposed fully-sampled Shells trajectory on the same 3.0 T MR system and the 8-channel receive-only brain coil. The fully-sampled Shells trajectory was designed based on the acquisition parameters shown in Table 1, with $\Delta x = 1$ mm³, FOV diameter = 24 cm, readout length $T = 512$, and BW = ±62.5 kHz. The acquisition of shells with different radii was arranged as depicted in Fig. 3, with the inner-most k-space shell acquired at TI = 900 ms. The data acquisition time window (1.1 s) and repetition time (TR = 2.3 s) of MP-Shells were chosen to match with that of a routine clinical MP-RAGE sequence with the same data acquisition settings. Fat saturation module (6 ms duration) was enabled for MP-Shells with $n = 3$, i.e., every three shells readouts were played following each fat saturation module, yielding a total of 115 shells readouts and 38 fat saturation modules in each TR. The total acquisition time for the MP-Shells was 6 min and 43 sec (Table 1). As a comparison, the same subject was also scanned using the clinical MP-RAGE sequence with the matching inversion time (TI) and data acquisition parameters as that for the MP-Shells (FOV diameter = 24 cm, $\Delta x = 1$ mm³ (isotropic), matrix size = 240³, BW = ±62.5 kHz, FA = 8°, repetition time (TR) = 2.3 sec, TI = 900 ms, TE = 1.8 ms, readout time = 4.7 ms, acquisition time = 9 min and 14 sec). A B_0 mapping sequence with a FOV matching the MP-Shells scan was performed using the same acquisition parameters as the B_0 mapping sequence used in the phantom scan.

The brain of a second healthy volunteer was also scanned under IRB-approved protocols using the proposed MP-Shells trajectory both without (i.e., fully-sampled) and with (i.e., undersampled) acceleration. Acquisition parameters similar to the previous brain scan were used except for the 1.1 mm isotropic resolution. Details of the acquisitions are shown in Table 1. For the accelerated MP-Shells, the Shells radius threshold in Eq. 5 was chosen as $n_0 = 72$ and 55, corresponding to an undersampling rate of 1.4 and 2.1, respectively.

Data Processing

All data processing was performed under a hybrid C++/Matlab environment (v 8.6) on a dual 8-core 2.6 GHz machine with memory of 128 GB. All NUFFT operators used in this work were implemented using a 5-point Kaiser-Bessel window and $1.25\times$ oversampling. The acquired k-space data were reconstructed onto a $240\times 240\times 240$ image grid with an isotropic 1.0 mm^3 reconstruction resolution. The phantom and *in vivo* data sets acquired with the fully sampled Shells trajectory were reconstructed using a non-iterative type-III NUFFT-based reconstruction framework that simultaneously accounts for the off-resonance effect and image gradient nonlinearity during reconstruction (34). The off-resonance effects in the phantom and *in vivo* data sets were corrected based on a time-segmentation method (35) with 16 or 8 segments for the phantom and *in vivo* scans, respectively. The off-resonance map was estimated based on the regularized regression model described by Funai et al. (40), which was adapted for multi-channel scenario and solved using a graph-cut based α -expansion procedure (41). For the fully sampled Shells trajectory, image reconstruction was performed on a coil-by-coil basis and subsequently root-sum-of-squares combined.

The phantom and *in vivo* images acquired using the accelerated Shells trajectory were reconstructed by solving the SENSE-type reconstruction problem defined in Eq. 6 via CG iteration (number of iterations = 5) with $\lambda = 1e-5$ for SPGR-style Shells and $\lambda = 1e-2$ for MP-Shells. The sensitivity maps of the phased array receive coils were estimated from B_0 calibration scan data using ESPIRiT (42) (calibration region of size 24×24 , kernel size of 6×6). For comparison, the fully sampled Shells images were also reconstructed using the same iterative solver. The operator $\mathbf{A}$ in Eq. 6 was implemented using a type-II NUFFT, with off-resonance effects accounted for within $\mathbf{A}$ using the time-segmentation approach (43). The reconstructed images were then discrete-Fourier-transformed into Cartesian k-space, and gradient-nonlinearity corrected using the type-I NUFFT operator, mirroring the approximation strategy previously derived for Cartesian SENSE acquisitions (44).

RESULTS

For the data acquisition setup used in the simulation and phantom experiments, a total of 120 shells and 20095 interleaves are designed to achieve a 1.0 mm^3 isotropic acquisition with a FOV of 24 cm. The total number of samples for Shells acquisition is approximately 74% of that for the Cartesian acquisition with the same acquisition parameters. Examples of the fully sampled inner ($n = 3$) and outer ($n = 100$) shells trajectories designed for the simulation and phantom experiments using the proposed method (Eqs. 1 and 2) are shown in Fig. 1. The number of interleaves for the outer shells (Fig. 1b) is 8-fold decimated to show details of each interleave. The inner shells such as that in Fig. 1a (shell radius index = 3) can be sampled using a single interleave, while the larger shells such as that in Fig. 2b (shell index = 100) require multiple interleaves (341 interleaves in this example) to achieve full sampling.

The 3D ACR phantom images acquired using the proposed fully sampled Shells trajectory are reformatted in axial, sagittal, and coronal planes, and shown in Fig. 4. The images are sinc-interpolated to 512×512 matrices for display. Table 1 lists details of the trajectory design and acquisition time. Fig. 4 shows that the images acquired using the proposed Shells

trajectory have minimal aliasing artifacts. The magnified panel in Fig. 4 next to the axial slice shows the resolution inserts of the ACR phantom. Each of the three inserts groups is made up of resolution dots corresponding to a resolution of (from top to bottom) 1.1, 1.0, or 0.9 mm, respectively. Figure 4 shows that the middle array (see red arrows) corresponding to the 1 mm resolution inserts can be well resolved, consistent with the imaging resolution setup of the trajectory design.

Examples of the ACR phantom images reconstructed using the SENSE framework from the fully sampled SPGR-Shells, and accelerated Shells acquisition with various acceleration factors are shown in Fig. 5. The images reconstructed using non-iterative gridding are also shown in supporting Figure S1 for comparison. The aliasing effects from undersampling shown in the gridding reconstructed images are largely reduced using the iterative SENSE reconstruction. As expected, data undersampling causes degraded image SNR in the reconstructed images. Also note the loss of low contrast detectability due to increased noise level when higher acceleration factors are applied, as denoted by the increased difficulty to resolve the low contrast inserts in the phantom (especially the smaller inserts).

The images acquired from a healthy volunteer's brain using the proposed MP-Shells, as well as the clinical MP-RAGE, are reformatted in three orthogonal planes and shown in Fig. 6. The fat tissues in the MP-Shells images, such as that close to the peripheral of skull, are largely suppressed by the fat saturation module embedded between different shells readouts (Figs. 6b,d,f), demonstrating the effectiveness of this strategy. Compared to the clinical MP-RAGE, the MP-Shells images exhibit slightly enhanced white matter to gray matter contrast. The signal to noise ratio (SNR) of white matter and gray matter, defined as the ratio between the mean signal intensity and the standard deviation in background, are measured from the region of interests (ROI) in gray, white matter, and background noise regions in different planes, and summarized in Table 2. Also shown in Table 2 are the gray-to-white matter contrast to noise ratios (CNR) of both image sets calculated as the difference between their mean signal intensities, divided by the background noise standard deviation. Compared with MP-RAGE, as shown in Table 2, the MP-Shells images features an increased SNR of white matter (MP-Shells: 50.86 ± 1.86 ; MP-RAGE: 45.39 ± 2.58), and similar gray matter SNR (MP-Shells: 30.17 ± 2.91 ; MP-RAGE: 30.17 ± 3.22), which contributes to an enhanced CNR between gray and white matter (MP-Shells: 20.69 ± 2.90 ; MP-RAGE: 15.21 ± 2.16). The improved SNR and CNR are also reflected in the better visualization of the dilated perivascular space in the MP-Shells image (marked as red arrows in Fig. 6).

Examples of images acquired on another healthy volunteer brain using the fully sampled and SENSE accelerated MP-Shells with a factor of 1.4 and 2.1 are shown in Fig. 7. Relatively moderate accelerations are used as an MP acquisition usually has lower SNR compared to an SPGR acquisition. Compared to the fully sampled case, the accelerated MP-Shells images share similar image contrast between gray and white matter but with increased noise level. Similar to the phantom images, the *in vivo* MP-Shells results demonstrate increased SNR loss when a higher acceleration factor is applied.

DISCUSSION

In this work, we present a new magnetization-prepared non-Cartesian MRI acquisition based on the Shells trajectory and demonstrate a fully automated trajectory and gradient waveform design. Compared to the previous trajectory design, no manual intervention is required for this new implementation. The proposed trajectory design was tested using a series of phantom experiment; the ACR phantom results confirm the absence of aliasing artifacts and demonstrate that the 1mm imaging resolution setting were successfully achieved.

Due to the corner skipping and use of the time-varying gradient, the acquisition time of the fully sampled MP-Shells is 27% shorter than the unaccelerated, conventional Cartesian MP-RAGE (6:43 vs 9:14, Table 1). The time-saving is lower than theoretically predicted 50% acceleration comparing the spherical k-space coverage of Shells versus the Cartesian cube, due to the requirement to satisfy the Nyquist sampling criteria at the equator of each shell, which leads to slight oversampling close to the two poles of each shell. This is similar to the case when setting Nyquist sampling for other non-Cartesian trajectories such as radial acquisition, which can cause oversampling in the center of k-space when Nyquist criteria is strictly followed (28). The built-in fat saturation in MP-Shells can successfully null the signal from the subcutaneous adipose tissue, which reduces the fat-induced blurring effects and increases conspicuity of brain tissues. The images acquired using MP-Shells demonstrate expected image quality and slightly higher or equivalent white and gray tissue SNR compared to MP-RAGE acquisition (Table 2). The CNR between the grey and white matter of brain in MP-Shells is also superior to that of the MP-RAGE. This can be attributed to the higher sampling efficiency of Shells trajectory in capturing the evolving gray-to-white matter contrast. Note that the proposed MP-Shells concentrates the acquisition of the inner 3D k-space shells around the prescribed inversion time to improve the desired image contrast.

Previous work (45) compared the effect of centric and sequential Cartesian view orderings on gray to white matter CNR for an MP-RAGE type acquisition. Its results showed that the centric view ordering can improve the CNR by 14% compared to the conventional sequential view ordering used in MP-RAGE. On the other hand, the MP-Shells offers a 36% CNR improvement.

MP-RAGE scans often require long acquisition times, especially when high resolution is required. Parallel imaging is sometimes used at the cost of reduced SNR. The compatibility of the proposed Shells trajectory design with SENSE-type parallel imaging reconstruction has also been demonstrated, which allows acquisition time and image quality to be actively traded off. As expected, a progressive reduction in SNR with increased scan acceleration was observed in our phantom SPGR-type Shells experiments (Fig. 5), as well as the *in vivo* MP-Shells images (Fig. 7). Compared to non-iterative gridding, model-based iterative reconstruction can help to reduce the artifacts and improve image quality at the cost of longer computational time (also see supporting Figure S1). The current CG iteration implemented under an OpenMP-parallelized C++ environment based on type-II NUFFT operators required 570 s (5 iterations; 2 fold acceleration) for reconstruction versus 36 s/coil for non-iterative gridding. We expect that further code optimization and/or use of high-

throughput computing hardware (e.g., GPGPU) can further reduce computation times. Different from parallel imaging in 3D Cartesian acquisition, where undersampling can be performed along each individual axis, reducing the number of interleaves in the Shells trajectory causes undersampling in all three axes, as suggested in Fig. 1. The noisy appearance shown in the accelerated MP-Shells images (Fig. 7) can be explained by the undersampling along the z-direction, since the 8-channel head coil lacks sufficient sensitivity variation along the z-direction, which leads to unresolved aliasing in the images. For Shells acquisition, the aliasing effect is naturally incoherent similar to the other non-Cartesian trajectories like radial, and the unresolved aliasing effect appears to be noise-like in the reconstructed image. We have compared the g-factor map of the 2.1-fold accelerated MP-Shells (46) with that of a 2-fold Cartesian SENSE (Supporting Figure S3) acquisition, which suggests the g-factor penalty is not the main contributor of the noisy appearance in this example. The parallel imaging performance of Shells could be further improved by using a high channel-count receive array with channel sensitivity profiles of sufficient variation in three dimensions (47).

The acquisition strategy proposed for MP-Shells can potentially be extended to different MR acquisitions such as contrast enhanced MR angiography (10,48), double inversion recovery (DIR) (49), or other applications requiring imaging of dynamically changing contrast. In the case of DIR, the acquisition of the inner k-space shells can be synchronized with the time point when both CSF and white matter are attenuated as desired. However, further study is required to investigate this possibility.

The Shells trajectory uses time-varying gradient waveforms, which are usually limited by the gradient system hardware constraints including the gradient amplitude and slew rate limits. Therefore, it can potentially benefit from gradient systems with higher performance such as the compact, asymmetric gradient developed for neural, musculoskeletal (MSK), and infant applications (50). The higher slew rate of this system can allow each individual interleaves to swirl faster with a higher ω_n , which reduces the number of interleaves (M_n) required to comply with the Nyquist sampling requirement (Eq. 2). An approximate estimation suggests a 12% reduction of acquisition time assuming the 700 T/m/sec slew rate of the system.

We speculate that the shading artifacts shown on the edge of the phantom images (Figs. 4 and 5) are due to the effects of gradient eddy current and gradient delay, which can cause the actually played-out trajectory to deviate from design, and were not corrected in these examples. Since these effects mainly affected the periphery of the FOV in these examples, we believe their impact on the resolution inserts (which is close to isocenter) is small. As with other non-Cartesian acquisitions, the Shells trajectory can benefit from correction of various system imperfections such as eddy current and gradient delay. We expect image quality to be further improved after appropriate gradient trajectory calibration using similar methods developed for other nonCartesian acquisitions (51–53).

CONCLUSIONS

In this work, we demonstrated a novel, fully-automated trajectory and gradient waveform design for the Shells pulse sequence. It was applied to a magnetization-prepared MRI acquisition in a fully-sampled mode, as well as with parallel imaging. The pulse sequence achieved efficient data acquisition with improved gray-to-white brain matter contrast. Phantom and volunteer results demonstrate the feasibility of the fully-automated trajectory and gradient waveform design, and its compatibility with parallel imaging.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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APPENDIX

Here, we show the derivation of the number of interleaves required for the n -th shell (Eq. 2) to satisfy the Nyquist requirement. The largest separation between two neighboring interleaves is observed on the equator of the k -space shell, as shown in Fig. 1c. The distance between two interleaves, l , can be expressed as:

$$s \approx l \sin \alpha = k_{r,n} \Delta \theta \sin \alpha = n \Delta k_r \Delta \theta \sin \alpha = n \Delta k_r \frac{2\pi}{M_n} \sin \alpha \quad (\text{A1})$$

where $\Delta \theta = \frac{2\pi}{M_n}$ denotes the difference of the initial phase, $\theta_{n,m}$, between two neighboring interleaves, $l = k_{r,n} \Delta \theta$ approximates the arc distance between these two interleaves in the transverse plane, and α is the tangent angle of the helical spiral interleave made with the transverse plane. α can be calculated as:

$$\alpha = \arctan \frac{dz}{dl} = \arctan \frac{\omega_z dt}{\omega_n dt} = \arctan \frac{\pi}{\omega_n} \quad (\text{A2})$$

where $\omega_z = \pi$ by definition (Eq. 1). Inserting Eq. A2 back to Eq. A1 yields:

$$s \approx n \Delta k_r \frac{2\pi}{M_n} \sin \left(\arctan \frac{\pi}{\omega_n} \right) \quad (\text{A3})$$

Taking into account the Nyquist sampling requirement ($s \leq \Delta k_r$), as well as the fact that M_n is an integer, Eq. A3 leads to:

$$M_n = \lceil 2\pi n \sin\left(\arctan\frac{\pi}{\omega_n}\right) \rceil$$

which gives the second part of Eq. 2.

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Figure 1.

Examples of 3D helical spiral interleave/readout designed for an inner and outer k-space shells (a, b). The inner shells close to the k-space center have small radii, and can be sampled using one single readout (a), while the outer shells with large radii (b) need more readouts (color-coded) to satisfy the Nyquist sampling requirement. Also refer to text for details. c: Illustration for the determination of number of interleaves/readouts required for each k-space shell to satisfy the Nyquist sampling requirement (see Appendix for detail).

Figure 2.

Flowchart showing the design steps to generate the gradient waveform for the Shells acquisition. A given set of data acquisition parameters (FOV, isotropic resolution, i.e., Δx) are first used to calculate the number of shells (N) and separation between shells (Δk_r). Parametrized trajectory is then calculated using Eq. 1 with number of sampling points T and initial value of $\omega_{n,0} = 2\pi^2 n$. Based on the pre-set BW, the Nyquist gradient limits are calculated (G_{Nyquist}). The minimum between the Nyquist gradient limit and the gradient hardware constraints (G_0, S_0 for maximum system gradient amplitude and slew rate), are used as the input for the automated gradient waveform generation method to yield the gradient waveform for the first interleave with $\theta_{n,0} = 0$. The angular rotational speed (ω_n) that gives the targeted number of sampling of points, T , is then determined. Finally, the number of interleaves M_n for each shell is calculated based on ω_n , which is then used to update the other interleaves with different $\theta_{n,m}$.

Figure 3.

(a) Evolution of white and gray matter magnetizations, as well as the gray to white matter contrast, following an inversion pulse in a MP-Shells repetition time (TR). The inversion time (TI), repetition time (TR), and length of acquisition window, recovery time are chosen to match with the clinical MP-RAGE sequence. The Shells readouts order within the acquisition window is arranged based on their k-space radii so that the inner most shell is acquired at TI time, and shells of smaller radii (inner shells) are preferentially placed closer to the TI time, as denoted by the colorbar, where the brighter shades represent the interleaves from the shells with smaller radii while the darker shades represent interleaves from the shells with larger radii. (b) Illustration for the arrangement of fat saturation and shells acquisition modules within acquisition window. As each acquisition module is short, the fat saturation is performed before every n shells readouts ($n = 3$ in this example) to suppress the fat components in images.

Figure 4.

Examples of a 3D image set acquired using the automatically designed fully-sampled SPGR-type Shells trajectory, reformatted in sagittal (a), coronal (b), and axial (c) planes. The magnified panel in (c) shows the three groups (31 resolution dots in each group) of resolution inserts of the ACR phantom, which are corresponding to a resolution of 1.1 mm (top), 1.0 mm (middle), and 0.9 mm (bottom), respectively. The 1.0 mm group (red arrow) can be successfully resolved, in consistent with the prescribed 1.0 mm resolution setting of this acquisition.

Figure 5. Examples of ACR phantom images (axial plane) acquired using the fully-sampled, SPGR-type Shells, and Shells with different acceleration. Images were reconstructed using the iterative, SENSE-type reconstruction solved via conjugate gradient (CG).

Figure 6.

Examples of 3D MP-RAGE (a,c,e) and MP-Shells (b,d,f) images acquired on a healthy volunteer. Images are reformatted in axial (a,b), coronal (c,d), and sagittal planes(e,f). The red arrows marked the improved visualization of the perivascular space in MP-Shells. The ROIs for white and gray matters used in SNR and CNR measurement are marked using green and red circles, respectively. Blue circles mark the ROIs for background noise standard deviation measurements.

Figure 7.

Examples of fully-sampled MP-Shells (a), and MP-Shells with 1.4-fold (b) and 2.1-fold (c) SENSE acceleration reformatted in axial plane.

Table 1

Details of Data Acquisition Settings.

Object	Sequence	SENSE Factor	FOV (cm)	Res (mm ³)	BW (kHz)	FA	# of Samples per RO	RO Time (ms)	# of RO	TR/TI for MP (sec)	# of RO per TR	# of TR	Acq. Time (min:sec)
ACR Phantom	SPGR-Shells	Fully-Sampled	24	1×1×1	±125	8°	512	5.9	20095	–	–	–	2:00
ACR Phantom	SPGR-Shells	2×	24	1×1×1	±125	8°	512	5.9	9866	–	–	–	0:59
ACR Phantom	SPGR-Shells	3×	24	1×1×1	±125	8°	512	5.9	6759	–	–	–	0:41
ACR Phantom	SPGR-Shells	4×	24	1×1×1	±125	8°	512	5.9	4972	–	–	–	0:30
Healthy Volunteer #1	MP-Shells	Fully-Sampled	24	1×1×1	±62.5	8°	512	7.8	19550	2.3/0.9	115	170	6:43
Healthy Volunteer #1	MP-RAGE	Fully-Sampled	24	1×1×1	±62.5	8°	240	4.7	240×240	2.3/0.9	240	240	9:14
Healthy Volunteer #2	MP-Shells	Fully-Sampled	24	1.1×1.1×1.1	±62.5	8°	512	7.8	14950	2.3/0.9	115	130	5:07
Healthy Volunteer #2	MP-Shells	1.4×	24	1.1×1.1×1.1	±62.5	8°	512	7.8	10350	2.3/0.9	115	90	3:35
Healthy Volunteer #2	MP-Shells	2.1×	24	1.1×1.1×1.1	±62.5	8°	512	7.8	6900	2.3/0.9	115	60	2:26

Note: FOV: diameter of the spherical imaging volume; Res: isotropic resolution; FA: flip angle; Acq. Time: acquisition time; # of RO: number of readouts/interleaves. The number of RO refers to the number of readouts/interleave for Shells acquisition, and refers to the number of phase encoding lines for MP-RAGE.

Table 2

Signal-to-noise ratios (SNR) of white and gray matter (WM, GM), as well as gray-to-white matter contrast-to-noise ratio (CNR), measured from the ROIs in MP-RAGE and MP-Shells images.

Sequence	SNR (WM) (mean $\pm$ std)	SNR (GM) (mean $\pm$ std)	CNR (GM to WM) (mean $\pm$ std)
MP-RAGE	45.39 $\pm$ 2.58	30.17 $\pm$ 3.22	15.21 $\pm$ 2.16
MP-Shells	50.86 $\pm$ 1.86	30.17 $\pm$ 2.91	20.69 $\pm$ 2.90