Published in final edited form as:

Phys Med Biol. ; 64(4): 04NT01. doi:10.1088/1361-6560/aafcc5.

Partial Fourier shells trajectory for non-Cartesian MRI

Shengzhen Tao¹, Yunhong Shu^{1,*}, Joshua D Trzasko¹, John Huston III¹, and Matt A Bernstein¹

¹Department of Radiology, Mayo Clinic, Rochester, MN, USA

Abstract

Non-Cartesian MRI acquisition has demonstrated various advantages in many clinical applications. The shells trajectory is a 3D non-Cartesian MRI acquisition technique that samples the k-space using a series of concentric shells to achieve efficient 3D isotropic acquisition. Partial Fourier acquisition is an acceleration technique that is widely used in Cartesian MRI. It exploits the conjugate symmetry of k-space measurement to reduce the number of k-space samples compared to full-k-space acquisition, without loss of spatial resolution. For a Cartesian MRI acquisition, the direction of partial Fourier acceleration is aligned either with the phase encoded or frequency encoded direction. In those cases, the underlying image matrix can be reconstructed from the undersampled k-space data using a non-iterative, homodyne reconstruction framework. However, designing a non-Cartesian acquisition trajectory that is compatible with non-iterative homodyne reconstruction is not nearly as straightforward as in the Cartesian case. One reason is the non-iterative homodyne reconstruction requires (slightly over) half of the k-space to be fully sampled. Since the direction of partial Fourier acceleration varies throughout the acquisition in the non-Cartesian trajectory, directly applying the same partial Fourier acquisition pattern (as in Cartesian acquisitions) to a non-Cartesian trajectory does not necessarily yield a continuous, physically-achievable trajectory. In this work, we develop an asymmetric shells trajectory with fully-automated trajectory and gradient waveform design to achieve partial Fourier acquisition for the shells trajectory. We then demonstrate a non-iterative image reconstruction framework for the proposed trajectory. Phantom and in vivo brain scans based on spoiled gradient echo (SPGR) shells and magnetization-prepared shells (MP-shells) were performed to test the proposed trajectory design and reconstruction method. Our phantom and in vivo results demonstrate that the proposed partial Fourier shells trajectory maintains the desirable image contrast and high sampling efficiency from the fully sampled shells, while further reducing data acquisition time.

Keywords

partial Fourier; shells trajectory; non-Cartesian MRI; homodyne reconstruction

*Please direct all correspondence to: Yunhong Shu, Ph.D., Mayo Clinic, 200 First Street SW, Rochester, MN 55905, USA, Phone: +1 507 255 3325, Fax: +1 507 284 9771, shu.yunhong@mayo.edu.

1. Introduction

The shells trajectory is a three-dimensional (3D) non-Cartesian MRI technique for efficient isotropic acquisition (Irrazabal and Nishimura, 1995; Shu *et al.*, 2006; Shu *et al.*, 2009; Zahneisen *et al.*, 2012; Shu *et al.*, 2018). It uses a series of concentric k-space shells to sample a 3D k-space, with each individual shell in turn sampled by a group of helical spiral interleaves traversing its surface from one pole to the other (Shu *et al.*, 2018). The shells trajectory offers several intrinsic advantages. By only acquiring a spherical k-space around the center, the shells trajectory possesses high data acquisition efficiency, and allows the essential k-space information to be captured without significantly compromising image resolution (Shu *et al.*, 2006). Furthermore, with the shells trajectory, the inner k-space shells (i.e., the ones with smaller radii) determine the image contrast, and can be efficiently sampled within several interleaves. 3D acquisitions can be designed to effectively capture time-evolving contrast (Fain *et al.*, 1999), and shells is well-suited for these applications. The acquisition order of shells readouts can be arranged so that the acquisition of inner-most shells is synchronized with the time when a certain transient image contrast needs to be captured. For example, in gadolinium-enhanced MR angiography, the inner shells can be acquired when the contrast agents cause peak enhancement of the arteries of interest, while suppressing the venous phase signal (Shu *et al.*, 2009). In a magnetization-prepared sequence, the inner k-space shells can be acquired when the peak contrast between the white and gray matters is achieved, following a magnetization-preparation pulse, as we demonstrated in (Shu *et al.*, 2018).

Partial Fourier acquisition is a widely-used acquisition acceleration technique in MRI that exploits the conjugate symmetry of the k-space signal, in cases where background phase field is spatially smooth and thus compact around the origin in k-space (Noll *et al.*, 1991b). This background phase field invariably contains some phase errors, which can be caused by system imperfections and imaging object dependent effects, such as hardware group delay, gradient eddy currents, B1 field inhomogeneity, patient motion, and susceptibility effect (Bernstein *et al.*, 2004). For a Cartesian MRI acquisition, the direction of partial Fourier acquisition is aligned with either a phase encoding or frequency encoding direction. In either of those cases, the underlying image matrix can be estimated from the undersampled k-space data using a non-iterative, homodyne reconstruction framework with a phase variation correction using the phase map estimated from a low resolution image obtained from the central k-space region (Noll *et al.*, 1991b). Compared to the alternative iterative partial Fourier reconstruction methods such as POCS (Cuppen and van Est, 1987; Lee *et al.*, 2007; Liang *et al.*, 1992), the non-iterative homodyne reconstruction has the advantage of higher computation efficiency, and has therefore gained wide popularity.

However, designing a non-Cartesian acquisition trajectory that is compatible with non-iterative homodyne reconstruction is not nearly as straightforward as in the Cartesian case. One reason is the non-iterative homodyne reconstruction requires (slightly over) half of the k-space to be fully sampled. Since the direction of partial Fourier acceleration varies throughout the acquisition in the non-Cartesian trajectory, directly applying the same partial Fourier acquisition pattern (as in Cartesian acquisitions) to a non-Cartesian trajectory does not necessarily yield a continuous, physically-achievable trajectory (Block and Frahm,

2005). For example, for 2D spiral acquisition, simply skipping part of the high-pass k-space region can break the continuity of spiral readout. Non-Cartesian partial Fourier acquisition methods based on different variable density spiral acquisition schemes have also been described. However, in these cases, an iterative reconstruction such as POCS is typically required for image reconstruction (Lee *et al.*, 2007; Moriguchi *et al.*, 2004).

Recently, we have described an automated shells trajectory design (Shu *et al.*, 2018), which allows a fully sampled shells trajectory and its gradient waveforms to be automatically determined for a prescribed acquisition setting. In this work, a new asymmetric 3D shells trajectory design is presented. It combines the advantage of the shells trajectory and the improved acquisition efficiency of the partial Fourier acquisition. A non-iterative homodyne reconstruction framework is developed to be used in conjunction with it. The proposed trajectory design and reconstruction approach are then tested using a series of phantom and in vivo data sets.

2. Materials and methods

2.1 Partial Fourier shells Trajectory

For a 3D MR acquisition with isotropic resolution (Δx) and field-of-view (FOV), the 3D k-space can be sampled using a series of concentric shells with a radius of $k_n = n\Delta k = \frac{n}{\text{FOV}}$, with $0 \leq n < N$ and $N = \frac{\text{FOV}}{2\Delta x}$ ensuring full k-space sampling along the radial direction. The n -th k-space shell can be sampled using a series of helical spiral interleaves with the following expression:

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

where ω_n denotes the azimuthal rotation speed of this interleave which varies for each shell and is therefore a function of n ; $\theta_{n,m} = \frac{2m\pi}{M_n}$ is the initial phase of the m -th interleaves ($0 \leq m < M_n$) with M_n denoting the number of interleaves required to satisfy Nyquist sampling along the k-space azimuthal direction for the n -th shell. The determination of ω_n and M_n will be discussed later.

The parameter $\tau \in [\tau_{\min}, \tau_{\max}]$ controls the sampling starting and ending points along the interleave direction. For a fully sampled shells acquisition, $\tau_{\min} = 0$ and $\tau_{\max} = 1$. In this case, the trajectory starts from the “north pole” of each shell ($k_{n,m,z}(\tau = 0) = k_n$) to the “south pole” ($k_{n,m,z}(\tau = 1) = -k_n$), as demonstrated in Figure 1. Since the inner k-space shells have smaller radii compared to the outer k-space shells, it typically requires only a few readouts to satisfy the Nyquist sampling requirement (Figure 1a vs. 1b).

To achieve partial Fourier acquisition, $\tau_{\max}$ can be modified as:

$$\tau_{\max, n} = \begin{cases} 1, & \text{if } n \leq (2\kappa - 1)N \\ \frac{1}{\pi} \cos^{-1} \left(\frac{(1 - 2\kappa)N}{n} \right), & \text{if } n > (2\kappa - 1)N \end{cases} \quad (2)$$

so that the trajectory samples from to $k_{n,m,z}(\tau = 0) = k_n$ to $k_{n,m,z}(\tau = \tau_{\max}) = -\frac{(2\kappa - 1)N}{\text{FOV}}$, where κ denotes the partial Fourier sampling factor with $0.5 < \kappa \leq 1$. An example of trajectory designed with $\kappa = 0.7$ is shown in Figure 1. For the partial Fourier shells, the trajectory for inner k-space shells (i.e., $n \leq (2\kappa - 1)N$) are the same as that of a fully sampled shells (Figure 1a vs. 1c). However, the outer shells are asymmetrically (partially) sampled, as shown in Figure 1d: the trajectories fully sample the upper k-space hemisphere along the k_z direction ($0 \leq k_z \leq \frac{N}{\text{FOV}}$) while only sample part of the lower hemisphere ($-\frac{(2\kappa - 1)N}{\text{FOV}} \leq k_z < 0$), hence achieving a partial Fourier acquisition along k_z with a factor of κ . When $\kappa = 1$, the trajectory reduces to the fully sampled shells trajectory. Also note that a fully sampled lower k-space along k_z ($-\frac{(2\kappa - 1)N}{\text{FOV}} \leq k_z \leq \frac{(2\kappa - 1)N}{\text{FOV}}$) is available based on this design, from which the residual phase deviation due to various effects, such as B1 field inhomogeneity and gradient delay and eddy currents, can be estimated.

2.2 Gradient Waveform Generation

The trajectory described by Equations. 1 and 2 is a continuous function of τ , while, on the other hand, the input waveform to the gradient system is typically a discrete vector. Furthermore, the trajectories shown in Eq. 1 depend on the parameters ω_n and M_n , which have not been determined up to this point. Denoting T as the user-specified constant representing the number of k-space samples along each interleave (e.g., $T = 512$ points per interleave/readout), the trajectory parameters (ω_n and M_n) and the readout gradient waveforms corresponding to each interleave can be generated using a fully automated iterative gradient design scheme, as we developed previously for the fully sampled shells trajectory (Shu *et al.*, 2018). In this work, this scheme is adapted to partial Fourier asymmetric shells trajectory.

Note that the first shells interleave ($m = 0$) of each k-space shell has an initial phase of $\theta_{n,m=0} = 0$ (and therefore is independent from M_n), and further that the m -th interleave on the same shells (and its gradient waveforms) can be obtained by rotating the first interleave azimuthally by an initial angle $\theta_{n,m}$. Consequently, the design of all the interleaves on each individual shell can start with the first interleave with zero initial azimuthal angle. For any given trajectory parameter including ω_n , partial Fourier factor (κ), number of samples per interleave (T), and acquisition bandwidth (BW), the gradient waveforms corresponding to this trajectory can be obtained using an automated design method described by Lustig *et al.* (Lustig *et al.*, 2008), which calculates the time-optimal gradient waveforms of three physical axes that sample along the input k-space trajectory while satisfying given gradient amplitude and slew rate limits. These gradient amplitude and slew rate limits are typically determined by gradient system hardware constraints as well as the Nyquist sampling requirement along

the interleave direction. Due to the time-optimal nature of this design method, the gradient waveforms generated using this method does not necessarily yield the same number of sample point (i.e., T) as specified. To ensure a total of fixed T points for each interleave, the gradient generation can be repeated by iteratively adjusting ω_n until a total of T samples are obtained (Shu *et al.*, 2018). Note that the trajectory parameter ω_n for a partial Fourier shells acquisition implicitly depends on the partial Fourier factor κ , i.e., $\omega_n = \omega_n(\kappa)$. This is because partial Fourier shells acquisition reduces the k-space volume that needs to be covered. For a given target readout sample point number T , a higher partial Fourier factor allows for a higher $\omega_n(\kappa)$ to be used (i.e., more azimuthal turns per spiral interleave) while still satisfying the same gradient hardware and slew rate limits. Examples of $\omega_n(\kappa)$ values for different κ (assuming typical brain scan acquisition parameters) are summarized in Table 1.

With $\omega_n(\kappa)$ determined, the number of interleaves ($M_n(\kappa)$) required for each k-space shell can be calculated by ensuring Nyquist sampling along the azimuthal direction. This is achieved by limiting the maximal k-space distance between two adjacent shells interleaves (usually on the “equator” of each shell) to satisfy the Nyquist sampling criterion, which gives:

$$M_n(\kappa) = \begin{cases} 1, & \text{if } \omega_n(\kappa) \geq 2\pi^2 n \\ \left\lceil 2\pi n \sin\left(\arctan\left(\frac{\pi}{\omega_n(\kappa)}\right)\right) \right\rceil, & \text{if } \omega_n(\kappa) < 2\pi^2 n \end{cases} \quad (3)$$

Note that different from the fully sampled shells, $M_n(\kappa)$ has κ -dependence which is inherited from $\omega_n(\kappa)$.

The dependence of $\omega_n(\kappa)$ and $M_n(\kappa)$ on the partial Fourier factor κ has several implications. First, given the same number of readout points T , a partial Fourier shells trajectory could produce a higher $\omega_n(\kappa)$ than the fully sampled shells. Figure 2 compares the fully sampled shells trajectory (Figure 2a) with the partial Fourier shells trajectory (Figure 2b, $\kappa = 0.7$) with the same readout length ($T = 512$). Examples of 3D helical spiral readouts highlighted in Figure 2a and 2b are projected into 2D ($k_z - k_y$) k-space plane and shown in Figure 2d. Note that the partial Fourier trajectory (purple curve) possesses a larger curvature than a fully sampled shells trajectory (black curve) due to a smaller k_z coverage and a higher $\omega_n(\kappa)$ it allows. According to Eq. 3, a higher $\omega_n(\kappa)$ reduces the number of interleaves ($M_n(\kappa)$) required to satisfy the Nyquist sampling along the azimuthal direction, and ultimately reduces the data acquisition time. Secondly, for a similar acquisition time (i.e., similar $\sum_{n=0}^N M_n$), the number of readout point can be reduced for a partial Fourier shells compared to a fully sampled shells while achieving the same resolution and FOV. Figure 2c shows another design for the partial Fourier trajectory with the same acceleration factor ($\kappa = 0.7$) but a shorter readout length ($T = 384$ in 2c vs. $T = 512$ in 2b). The shorter readout enabled via partial Fourier shells can be used to reduce the off-resonance effects. Off-resonance refers to any signal that has a frequency offset from the Larmor frequency expected for a nucleus (e.g., water) located in a uniform main magnetic field and ideal linear gradients.

Similar to the other non-Cartesian trajectories, off-resonance spins in shells trajectories can introduce image blurring, spatial shift and distortion, as well as signal loss, as shown in the examples later.

2.3 Reconstruction for Partial Fourier shells

In this section, we demonstrate the compatibility of the proposed partial Fourier shells trajectory with non-iterative homodyne reconstruction. We adopt a coil-by-coil signal model for this purpose and derive the corresponding reconstruction scheme. Such a signal model can be applied to scenarios when imaging with a multi-channel phased array or with a single-channel coil, such as transmit/receive (TR) coil. The proposed trajectory design is especially relevant in the latter case, where conventional parallel imaging reconstruction methods (e.g., SENSE) are inapplicable. We note that, when multi-channel phase array receivers are utilized, data acquired using the proposed partial Fourier shells trajectory can also be reconstructed using conventional non-Cartesian parallel imaging reconstruction methods, e.g., (Pruessmann et al., 2001); however, for the reason mentioned above, this option will not be explored in this work.

Denoting $u(\mathbf{x})$ as the underlying real-valued imaging object, the κ -th MR k-space measurement of the c -th channel of a multi-channel receive array, $\mathbf{g}_c(\kappa)$, can be modeled using the following expression (Fessler, 2010):

$$\begin{aligned}\mathbf{g}_c[\kappa] &= \int_{\mathbf{x} \in \Omega} \left\{ u(\mathbf{x}) \left[s_c(\mathbf{x}) e^{-2\pi j \phi_c(\mathbf{x})} e^{-2\pi j (w(\mathbf{x}) \kappa \Delta t + \phi_0)} e^{-2\pi j \mathbf{k}[\kappa] \mathbf{x}} d\mathbf{x} \right] \right\} + \mathbf{n}_c[\kappa] \quad (4) \\ &= \int_{\mathbf{x} \in \Omega} \left\{ u_c(\mathbf{x}) e^{-2\pi j (\phi_c(\mathbf{x}) + \phi_0)} e^{-2\pi j (w(\mathbf{x}) \kappa \Delta t)} e^{-2\pi j \mathbf{k}[\kappa] \mathbf{x}} d\mathbf{x} \right\} + \mathbf{n}_c[\kappa]\end{aligned}$$

where $\mathbf{k}[\kappa]$ denotes the 3D k-space sampling positions using the proposed partial Fourier shells trajectory; $s_c(\mathbf{x})$ and $e^{-2\pi j \phi_c(\mathbf{x})}$ are the magnitude and phase distributions of the sensitivity profile of the c -th receive channel; $u_c(\mathbf{x}) = u(\mathbf{x}) s_c(\mathbf{x})$ denotes the image object function weighted by the magnitude of the sensitivity profile, and is also real-valued; $e^{-2\pi j (w(\mathbf{x}) \kappa \Delta t + \phi_0)}$ represent the off-resonance induced phase distribution with $w(\mathbf{x})$ denoting the off-resonance map which can be estimated from a separate calibration scan, t the sampling interval, and ϕ_0 a phase offset term; finally, $\mathbf{n}_c[\kappa]$ is the zero-mean Gaussian noise within the data samples. Note that when $c = 1$, Eq. 5 reduces to the case when imaging with single-channel receiver.

Using finite series approximation and Dirac delta pixel model (Fessler, 2010), i.e., $\mathbf{u}_c[n] \approx \sum_{\mathbf{n} \in \Theta} u_c(\mathbf{r}(\mathbf{n})) \delta(\mathbf{x} - \mathbf{r}(\mathbf{n}))$, the signal model in Eq. 4 can be reformatted into a linear algebraic form:

$$\mathbf{g}_c = \mathbf{A} \Psi_c \mathbf{u}_c + \mathbf{n}_c \quad (5)$$

where $\mathbf{A}$ represents the forward spatial encoding operator (N_k by N_u) including the off-resonance effect, with $\mathbf{A}[n, \kappa] = e^{-2\pi j \mathbf{k}[\kappa] \mathbf{r}[n]} e^{-2\pi j \mathbf{w}(\mathbf{r}[n]) \kappa \Delta t}$; $\mathbf{g}_c$ is the signal measurement (N_k by 1 vector) of the c -th receive channel; $\mathbf{u}_c$ denotes the real-valued image object (N_u by 1) weighted by the magnitude of the sensitivity profile corresponding to the c -th receive channel; N_k , N_u and C denote the number of k-space samples per channel, the number of image pixels, and the total number of receive channels, respectively; Ψ_c is a diagonal matrix (N_u by N_u) that approximates the spatial phase signal due to the phase variation of the receive phased array and the residual phase unaccounted for in the forward operator ($\Psi_c[n, n] = e^{-2\pi j(\phi_c(\mathbf{r}[n]) + \phi_0)}$), such as the gradient delay and eddy current induced phase.

Similar to the Cartesian-based, partial Fourier homodyne reconstruction, the image acquired using the proposed partial Fourier shells acquisition can be reconstructed based on the method-of-moments estimator on a channel by channel basis (Noll *et al.*, 1991b; Tao *et al.*, 2016), which yields $E\{\mathbf{g}_c\} = E\{\mathbf{A}\Psi_c\mathbf{u}_c + \mathbf{n}_c\} = \mathbf{A}\Psi_c\mathbf{u}_c$, with $E\{\cdot\}$ representing the expected values. It then follows that the real-valued image of each receive channel can be reconstructed as:

$$\mathbf{u}_c = \text{real}\{\Psi_c^* \mathbf{A}^* (\mathbf{A}\mathbf{A}^*)^{-1} \mathbf{g}_c\} \approx \text{real}\{\Psi_c^* \mathbf{A}^* \mathbf{D} \mathbf{g}_c\} \quad (6)$$

where $\mathbf{D} \approx (\mathbf{A}\mathbf{A}^*)^{-1}$ denotes the sampling density compensation function, which can be determined for the shells trajectory using the method described by Pipe and Menon (Pipe and Menon, 1999). Let Φ_L be a binary diagonal operator extracting the low-pass regions along the k_z axis sampled using the partial Fourier shells trajectory (i.e., the regions corresponding to $-\frac{(1-2\kappa)N}{\text{FOV}} \leq k_z \leq \frac{(1-2\kappa)N}{\text{FOV}}$), and let Φ_{H_1} and Φ_{H_2} be operators extracting the high-pass regions of the k-space along the partial Fourier acceleration direction that is sampled ($\frac{(1-2\kappa)N}{\text{FOV}} < k_z \leq \frac{N}{\text{FOV}}$) and not sampled ($\frac{N}{\text{FOV}} \leq k_z < \frac{(1-2\kappa)N}{\text{FOV}}$) using shells trajectory, respectively. Equation 5 can then be written as:

$$\begin{aligned} \mathbf{u}_c &\approx \text{real}\{\Psi_c^* \mathbf{A}^* \mathbf{D} \mathbf{g}_c\} \\ &= \text{real}\{\Psi_c^* \mathbf{A}^* \mathbf{D} (\Phi_{H_1} + \Phi_{H_2} + \Phi_L) \mathbf{g}_c\} \\ &= \text{real}\{\Psi_c^* \mathbf{A}^* \mathbf{D} (2\Phi_{H_1} + \Phi_L) \mathbf{g}_c\} + \text{real}\{\Psi_c^* \mathbf{A}^* \mathbf{D} (2\Phi_{H_2} - \Phi_{H_1}) \mathbf{g}_c\} \end{aligned} \quad (7)$$

Figure 3 shows an example of a partial Fourier shells trajectory that can be used for homodyne reconstruction. The k-space regions physically sampled during data acquisition (i.e., $\Phi_{H_1} + \Phi_L$) is shown in Figure 3a, while Figure 3b shows the region that is not acquired but implicitly assumed in the homodyne reconstruction (i.e., Φ_{H_2}).

Assuming that the underlying image object is real-valued and that the image phase is spatially smooth (which is an approximation for a single-channel TR coil or a low-count receiver), the k-space measurements should have conjugate symmetry, which yields:

$\text{real}\left\{\Psi_c^* \mathbf{A}^* \mathbf{D} \left(\Phi_{H_2} - \Phi_{H_1} \right) \mathbf{g}_c\right\} \approx 0$. Eq. 6 can therefore be simplified to:

$$\mathbf{u}_c \approx \text{real}\left\{\Psi_c^* \mathbf{A}^* \mathbf{D} \left(2\Phi_{H_2} - \Phi_{H_1} \right) \mathbf{g}_c\right\} \quad (8)$$

which suggests that each individual channel image can be reconstructed non-iteratively by applying a gridding reconstruction with density compensation ($\mathbf{A}^* \mathbf{D}$) to the “high-pass” filtered k-space data $\left(2\Phi_{H_2} - \Phi_{H_1} \right) \mathbf{g}_c$, followed by a phase variation correction using Ψ_c^* .

The phase map Ψ_c^* can be estimated using the fully sampled, low-pass region k-space data as:

$$\Psi_c = \text{diag}\left\{e^{j\angle\left\{\mathbf{A}^* \mathbf{D} \Phi_L \mathbf{g}_c\right\}}\right\} \quad (9)$$

with $\angle\{\cdot\}$ calculating the phase of a complex vector. The reconstruction formula in Eq. 9 is similar to the Cartesian homodyne reconstruction (Noll *et al.*, 1991b; Tao *et al.*, 2016). Note that the derivation in Equations 6 and 7 implicitly assumes that the k-space trajectory within the region Φ_{H_1} and Φ_{H_2} have conjugate symmetry (Noll *et al.*, 1991b), as shown in Figure

3.

2.4 Phantom Experiments

The proposed partial Fourier shells trajectory was implemented on a 3.0 T GE Signa whole-body MRI system (HDxt 16.0) with gradient amplitude and slew rate limit of 40 mT/m and 150 T/m/sec, respectively. The American College of Radiology (ACR) MRI quality control phantom was first scanned in an 8-channel receive-only brain coil (In Vivo, Gainesville, FL, USA) using the fully sampled and partial Fourier accelerated shells trajectories (with partial Fourier factor $\kappa = 0.6$ and 0.7) designed using the acquisition setting of a typical brain scan (spherical FOV = 24 cm, isotropic resolution = 1 mm, readout length $T = 512$, also see in Table 2). The partial Fourier shells with a factor of 0.6 was also designed with a reduced readout length ($T = 384$ vs. 512 , see Table 2). The shells interleaves of different shells were played in a center-out fashion with the inner-most shells' interleaves played first, followed by the outer k-space shells. Each shell interleaf was played following a small flip angle RF pulse in a manner similar to the spoiled gradient echo (SPGR) sequence. A separate dual-echo calibration scan with the following parameters was performed on the same imaging volume to estimate the off-resonance map (FOV = 24 cm³, acquisition matrix = 256×128×64, flip angle = 10°, echo spacing = 2.3 ms, TR = 9 ms, BW = ±31.25 kHz). All raw data were retained for offline processing.

To demonstrate the effect of shorter readout time compared to the longer readout time for PF-accelerated shells trajectory in the presence of off-resonance, the point spread function was calculated using numerical simulation for the partial Fourier shells (partial Fourier factor $\kappa = 0.7$, RO = 512) with and without off-resonance of 200/150 Hz, as well as the partial Fourier shells of shorter readout (partial Fourier factor $\kappa = 0.7$, RO = 384) and the off-resonance frequency of 200 Hz.

2.5 In vivo Experiments

A healthy volunteer brain was scanned under an IRB-approved protocol following written consent using the 8-channel receive-only brain coil on the same 3.0 T MR system. The fully sampled and the proposed partial Fourier shells trajectories with a partial Fourier factor $\kappa = 0.6$ and two readout lengths ($T = 512$ and 256) were designed assuming a typical brain scan acquisition setup (see Table 2 for detailed acquisition parameters). The magnetization prepared shells (MP-shells) were used in this experiment (Shu *et al.*, 2018). For an MP-shells acquisition, each TR consisted of a magnetization preparation pulse, data acquisition phase, and recovery phase, similar to a conventional Cartesian MP-RAGE acquisition (Mugler and Brookeman, 1990). Following an 180° magnetization preparation RF pulse, a series of shells readouts are performed during the data acquisition phase, as detailed in a previous work (Shu *et al.*, 2018). The order of shells interleaves was determined so that the acquisition starts from the outer k-space shells at the beginning of the acquisition phase in each TR. It then gradually transits to the inner k-space shells as the acquisition approaches the user-prescribed inversion time (TI), with the inner most shell acquired at TI time. The acquisition then gradually transits back to the outer k-space shells as it departs the inversion time. This arrangement is conceptually similar to a spiral-in/spiral-out readout fashion for a spiral acquisition (Glover and Law, 2001). The acquisition parameters were chosen to match with that of the clinical MP-RAGE acquisition with the same resolution and FOV setup (TR = 2.3 s, TI = 900 ms, and acquisition phase = 1.1 s). To reduce the blurring effect from the fat tissue, a fat saturation module was embedded in-between every three interleaves. The effectiveness of this strategy in fat saturation was demonstrated in a previous work (Shu *et al.*, 2018). A 3D dual echo GRE sequence was performed on the same imaging volume to estimate the off-resonance map (FOV = 24 cm^3 , acquisition matrix = $256 \times 128 \times 64$, flip angle = 10° , echo spacing = 2.3 ms, TR = 9 ms, BW = $\pm 31.25 \text{ kHz}$). As a comparison, a clinical Cartesian MP-RAGE sequence with parameters matching with that of the MP-shells is also performed on the same subject. The acquisition time of different sequences are shown in Table 2.

2.6 Data Processing

All data processing was performed on a dual CPU (8-core, 2.6 GHz) computer with 128 GB memory. The reconstruction framework was implemented in C++ environment with OpenMP-based parallelization. The operator $\mathbf{A}^*$ in Equations 8 and 9 can be implemented with the non-uniform fast Fourier transform (NUFFT) operators of either type II (Beatty *et al.*, 2005; Fessler and Sutton, 2003) or type III (Tao *et al.*, 2015b). In this work, we implemented $\mathbf{A}^*$ using the adjoint type-III NUFFT operator ($1.25 \times$ oversampled FFT and 5-point Kaiser-Bessel interpolation kernel) which simultaneously accounts for the off-resonance and gradient nonlinearity effects (Tao *et al.*, 2015b). All the images were

reconstructed onto a $240 \times 240 \times 240$ image grid with 1 mm^3 isotropic reconstruction resolution. Off-resonance correction was performed based on the time-segmented method (Noll *et al.*, 1991a) (26 segments for the phantom data set and 8 segments in vivo data sets, respectively). The off-resonance maps were estimated from a separate Cartesian off-resonance mapping scan based on a regularized regression model (Funai *et al.*, 2008) with multi-receiver extension, which was solved using a graph-cut based α -expansion method (Kolmogorov and Zabih, 2004). The images were reconstructed coil-by-coil using Equation 8 with phase map estimated from Equation 9. A Tukey window was applied to the weighting function ($2\Phi_{H_1} + \Phi_L$) to reduce the ringing effect due to the sudden transition of the

weighting function, similar to the Cartesian partial Fourier homodyne reconstruction (Bernstein *et al.*, 2004). The images of different receive channels were then root-sum-of-squares combined. The point spread function data were reconstructed using type-II NUFFT operator ($1.25 \times$ oversampled FFT and 5-point Kaiser-Bessel interpolation kernel) without off-resonance correction to observe the effect of off-resonance on partial Fourier shells design with different readout lengths. The MP-RAGE images were reconstructed using a type-I NUFFT operator ($1.25 \times$ oversampled FFT and 5-point Kaiser-Bessel interpolation kernel) which accounts for the gradient nonlinearity effect in data acquisition (Tao *et al.*, 2015a).

3. Results

The 3D ACR phantom image sets acquired using the fully sampled and partial Fourier SPGR-type shells with acceleration factor $\kappa = 0.7$ and 0.6 (readout length = 512) are reconstructed using the proposed non-iterative reconstruction and reformatted into sagittal planes and shown in Figure 4a to c. The images reconstructed from zero-filling the missing k-space samples for the partial Fourier shells are also shown for reference (Figure 4d and e). Figure 4 clearly demonstrates the blurring effect in the zero-filling reconstructions along the z-axis (Figure 4d and e). The image details are recovered using the proposed homodyne reconstruction (Figure 4b and c), compared to the fully sampled shells (Figure 4a). The line profiles across the z-axis for different reconstruction cases are also shown in Figure 4, which confirms the blurring effect in the zero filling cases and similar sharpness between the fully sampled and the proposed partial Fourier shells. Table 2 shows the acquisition times for the partial Fourier shells were reduced to 70% ($\kappa = 0.7$) and 58% ($\kappa = 0.6$) of the fully samples shells, respectively. The reconstruction time was 48 sec per coil for the fully sampled Shell, and 46 sec per coil for the partial Fourier shells (PF factor = 0.7) using homodyne reconstruction.

To demonstrate the effect of time-segmented NUFFT for off-resonance correction, Figure 5 shows the phantom image before and after off-resonance correction. Off-resonance causes image shading and distortion as pointed out by the arrows in (a), which is corrected after ORC (b).

Figure 6 shows the resolution inserts slice of the ACR phantom image sets acquired using the fully sampled (Figure 6a) and partial Fourier shells with readout lengths of 512 (Figure 6b) and 384 (Figure 6c). The resolution insert group corresponding to the 1.0 mm resolution

(middle group, red arrows) was well resolved in both the fully sampled and partial Fourier shells, confirming that the nominal resolution of 1.0 mm of trajectory design was reached. The partial Fourier shells with a shorter readout (Figure 6c vs. a and b) was able to reduce the residual off-resonance effect that is not compensated (see green arrows), while sacrificing acquisition efficiency. Note the increase of acquisition time for the partial Fourier shells with a shorter readout (2 min 2 sec) compared to that of the longer readout (1 min 24 sec) of the same acceleration factor ($\kappa = 0.7$), which is explained by the lower ω_n (reduced curvature) for the former case (Figure 2d).

Figure 7 compares the point spread function (PSF) along z direction (along which partial Fourier acceleration was applied) generated using partial Fourier shells of readout = 512 and 384 with 200 Hz off-resonance. Note that off-resonance correction was not performed to show its effect on shells with different readouts. The PSFs were sinc-interpolated to show details. The off-resonance causes a shift and distortion of PSF. The distortion is reduced for shells with a shorter readout of 384 compared to that of 512 points. The PSF generated using shells with readout = 512 and a lower off-resonance of 150 Hz is also shown. Note that its PSF is comparable to that of shells with readout = 384 and 200 Hz off-resonance. This is expected as the ratio between readout length ($384/512 = 0.75$) matched with that of off-resonance value ($150/200 = 0.75$).

Examples of *in vivo* brain images acquired using the fully sampled and partial Fourier accelerated MP-shells (acceleration factor $\kappa = 0.6$) are shown in Figure 8. The magnified insets are also shown. The clinical Cartesian MP-RAGE images are also shown for comparison. The partial Fourier MP-shells (Figure 8c and d) share similar image contrast compared to the fully sampled MP-shells (Figure 8b). As expected, Figure 8c and d also show an increased image noise level. The image noise levels for different cases were measured from the background regions and summarized in Table 3 for comparison, which shows that the image noise is increased by around 25% for partial-Fourier shells compared to the fully sampled shells. The contrast-to-noise ratio between the white and gray matter was measured based on the ROI shown in Figure 8 as the ratio between the mean intensity difference of the white/gray matter and the background standard deviation, and were 14.59 for MP-RAGE, 23.46 for fully sampled MP-shells, 20.75 for partial Fourier shells (readout length = 512), and 20.61 (readout length = 384), respectively. The improved white-to-gray matter CNR compared to MP-RAGE is in consistent with that reported in a previous work (Shu *et al.*, 2018). The image acquired with a shorter readout length (Figure 8d) is similar to that of a longer readout (Figure 8c) due to less prominent off-resonance effect and the fat-saturation that was used in *in vivo* brain acquisitions. Figure 9 shows the coronal reformats of different cases. The partial Fourier shells with shorter readout (d) demonstrates reduced susceptibility-induced distortion/blurring effect around the base of brain (arrows) compared to MP-shells with longer readout (b, c). The acquisition time was reduced from 9 min 14 sec of MP-RAGE to 6 min 43 sec for fully sampled MP-shells, and 4 min 25 sec for partial Fourier MP-shells (readout length = 512).

4. Discussion

In this work, we described an asymmetric, partial Fourier shells trajectory design, and demonstrated a non-iterative homodyne reconstruction framework for it. The proposed partial Fourier shells design is fully-automated. For prescribed acquisition setup, the acquisition trajectory and corresponding gradient waveforms can be automatically generated using the framework described in this work. The proposed asymmetric shells design also shares the advantage of fully-sampled shells in optimally capturing the time-evolving imaging contrast. As shown in Figure 6, the partial Fourier MP-shells share similar high contrast between white and gray matter compared to standard, Cartesian MP-RAGE. Compared to the fully sampled shells, the proposed partial Fourier shells can further increase data acquisition efficiency. Similarly to Cartesian partial Fourier acquisitions, the price for this increased acquisition speed is correspondingly increased image noise. The partial Fourier shells also allows a shorter readout length, which can be used to reduce the residual off-resonance effects due to imperfect off-resonance correction especially in regions with strong susceptibility (Figure 4). Note that the shorter readout length is not always possible for fully sampled shells due to insufficient gradient amplitude and slew rate for the trajectory to traverse from one pole to the other, or maximum bandwidth limitations of the receiver. We did not observe a substantial difference in the *in vivo* MP-shells results, due to the use of fat-saturation and relatively low off-resonance effect *in vivo*. For subjects with substantial dental work or metallic devices such as auditory brain implants (ABI), we speculate the improvement will be more pronounced. This is an area for future study.

We have applied the proposed partial Fourier shells trajectory in SPGR-type and MP-type acquisitions, but the proposed method can be further generalized to other applications such as contrast-enhanced MR angiography to efficiently capture the arterial phase signal while minimizing venous contamination (Hu *et al.*, 2004; Shu *et al.*, 2009). The high data acquisition efficiency of partial Fourier shells could also be helpful for 3D arterial spin labeling based MR perfusion imaging (Alsop *et al.*, 2015).

In this work, we have demonstrated a non-iterative reconstruction for the proposed partial Fourier shells trajectory. The compatibility with non-iterative homodyne reconstruction is a key intended feature of the proposed trajectory design. Consequently, the proposed trajectory can be applied to either the scenario of imaging using a multi-receiver phased-array, or using single-channel coil to accelerate data acquisition. Note that, in the latter case, conventional parallel imaging reconstruction methods (e.g., SENSE) are not applicable. As noted earlier, when data is acquired using a multi-channel phased array receiver and the proposed the proposed partial Fourier shells trajectory, it can – like standard shells data, and regardless of the degree of acquisition acceleration/undersampling performed – also be reconstructed using conventional non-Cartesian parallel imaging reconstruction methods like SENSE (Pruessmann *et al.*, 2001) and GRAPPA (Seiberlich *et al.*, 2007), including in combination with partial Fourier methods like homodyne, e.g., (King, 2000), or virtual conjugate coils (VCC) (Blaimer *et al.*, 2009). Such methods may offer improved reconstruction performance, but are expected to require more computational effort compared to the non-iterative coil-by-coil setup proposed here.

5. Conclusion

We developed an asymmetric partial Fourier shells trajectory and demonstrated its non-iterative homodyne reconstruction. The proposed trajectory inherits the high k-space sampling efficiency from the fully sampled shells trajectory, while further reducing data acquisition time or sensitivity to off-resonance effects.

Acknowledgement

This project was partly supported by grants R01EB010065, U01-EB024450, 1C06RR18898-01 from the National Institutes of Health.

REFERENCES:

- Alsop DC, Detre JA, Golay X, Gunther M, Hendrikse J, Hernandez-Garcia L, Lu H, MacIntosh BJ, Parkes LM, Smits M, van Osch MJ, Wang DJ, Wong EC and Zaharchuk G 2015 Recommended implementation of arterial spin-labeled perfusion MRI for clinical applications: A consensus of the ISMRM perfusion study group and the European consortium for ASL in dementia *Magn Reson Med* 73 102–16 [PubMed: 24715426]
- Beatty PJ, Nishimura DG and Pauly JM 2005 Rapid gridding reconstruction with a minimal oversampling ratio *IEEE Trans Med Imaging* 24 799–808 [PubMed: 15959939]
- Bernstein MA, King KF and Zhou XJ 2004 *Handbook of MRI pulse sequences* (Burlington, MA: Elsevier Academic Press)
- Blaimer M, Gutberlet M, Kellman P, Breuer FA, Kostler H and Griswold MA 2009 Virtual coil concept for improved parallel MRI employing conjugate symmetric signals *Magn Reson Med* 61 93–102 [PubMed: 19097211]
- Block KT and Frahm J 2005 Spiral imaging: a critical appraisal *J Magn Reson Imaging* 21 657–68 [PubMed: 15906329]
- Cuppen J and van Est A 1987 Reducing MR imaging time by one-sided reconstruction *Magn Reson Imaging* 5 526–7
- Fain SB, Riederer SJ, Bernstein MA and Huston J 3rd 1999 Theoretical limits of spatial resolution in elliptical-centric contrast-enhanced 3D-MRA *Magn Reson Med* 42 1106–16 [PubMed: 10571932]
- Fessler JA 2010 Model-Based Image Reconstruction for Mri *IEEE Signal Process Mag* 27 81–9 [PubMed: 21135916]
- Fessler JA and Sutton BP 2003 Nonuniform fast Fourier transforms using min-max interpolation *IEEE Trans Signal Proces* 51 560–74
- Funai AK, Fessler JA, Yeo DTB, Olafsson VT and Noll DC 2008 Regularized field map estimation in MRI *IEEE Trans Med Imaging* 27 1484–94 [PubMed: 18815100]
- Glover GH and Law CS 2001 Spiral-in/out BOLD fMRI for increased SNR and reduced susceptibility artifacts *Magn Reson Med* 46 515–22 [PubMed: 11550244]
- Hu HH, Madhuranthakam AJ, Kruger DG, Huston J 3rd and Riederer SJ 2004 Improved venous suppression and spatial resolution with SENSE in elliptical centric 3D contrast-enhanced MR angiography *Magn Reson Med* 52 761–5 [PubMed: 15389954]
- Irrarazabal P and Nishimura DG 1995 Fast three dimensional magnetic resonance imaging *Magn Reson Med* 33 656–62 [PubMed: 7596269]
- King KF. SENSE with partial Fourier homodyne reconstruction. *Proceedings of the 8th Annual Meeting of the ISMRM; Denver, Colorado: 2000.* 153
- Kolmogorov V and Zabih R 2004 What energy functions can be minimized via graph cuts? *IEEE Trans Pattern Anal* 26 147–59
- Lee JH, Pauly JM and Nishimura DG 2007 Partial k-space reconstruction for radial k-space trajectories in magnetic resonance imaging
- Liang Z, Boada F, Constable R, Haacke E, Lauterbur P and Smith M 1992 Constrained reconstruction methods in MR imaging *Rev Magn Reson Med* 4 67–185

- Lustig M, Kim SJ and Pauly JM 2008 A fast method for designing time-optimal gradient waveforms for arbitrary k-space trajectories IEEE Trans Med Imaging 27 866–73 [PubMed: 18541493]
- Moriguchi H, Lewin J and Duerk J Partial Fourier spiral reconstruction under consideration of off-resonance effects Proceedings of the 12th Annual Meeting of the ISMRM, (Kyoto, Japan, 2004), vol. Series) p 346
- Mugler JP 3rd and Brookeman JR 1990 Three-dimensional magnetization-prepared rapid gradient-echo imaging (3D MP RAGE) Magn Reson Med 15 152–7 [PubMed: 2374495]
- Noll DC, Meyer CH, Pauly JM, Nishimura DG and Macovski A 1991a A Homogeneity Correction Method for Magnetic-Resonance-Imaging with Time-Varying Gradients IEEE Trans Med Imaging 10 629–37 [PubMed: 18222870]
- Noll DC, Nishimura DG and Macovski A 1991b Homodyne detection in magnetic resonance imaging IEEE Trans Med Imaging 10 154–63 [PubMed: 18222812]
- Pipe JG and Menon P 1999 Sampling density compensation in MRI: Rationale and an iterative numerical solution Magn Reson Med 41 179–86 [PubMed: 10025627]
- Pruessmann KP, Weiger M, Bornert P and Boesiger P 2001 Advances in sensitivity encoding with arbitrary k-space trajectories Magn Reson Med 46 638–51 [PubMed: 11590639]
- Seiberlich N, Breuer FA, Blaimer M, Barkauskas K, Jakob PM and Griswold MA 2007 Non-Cartesian data reconstruction using GRAPPA operator gridding (GROG) Magn Reson Med 58 1257–65 [PubMed: 17969027]
- Shu Y, Bernstein MA, Huston J 3rd and Rettmann D 2009 Contrast-enhanced intracranial magnetic resonance angiography with a spherical shells trajectory and online gridding reconstruction J Magn Reson Imaging 30 1101–9 [PubMed: 19856444]
- Shu Y, Riederer SJ and Bernstein MA 2006 Three-dimensional MRI with an undersampled spherical shells trajectory Magn Reson Med 56 553–62 [PubMed: 16894580]
- Shu Y, Tao S, Trzasko JD, Huston J 3rd, Weavers PT and Bernstein MA 2018 Magnetization-prepared shells trajectory with automated gradient waveform design Magn Reson Med 79 2024–35 [PubMed: 28833440]
- Tao S, Trzasko JD, Shu Y, Huston J 3rd and Bernstein MA 2015a Integrated image reconstruction and gradient nonlinearity correction Magn Reson Med 74 1019–31 [PubMed: 25298258]
- Tao S, Trzasko JD, Shu Y, Weavers PT, Huston J 3rd, Gray EM and Bernstein MA 2016 Partial fourier and parallel MR image reconstruction with integrated gradient nonlinearity correction Magn Reson Med 75 2534–44 [PubMed: 26183425]
- Tao S, Trzasko JD, Shu YH, Huston J, Johnson KM, Weavers PT, Gray EM and Bernstein MA 2015b NonCartesian MR image reconstruction with integrated gradient nonlinearity correction Med Phys 42 7190–201 [PubMed: 26632073]
- Zahneisen B, Hugger T, Lee KJ, LeVan P, Reisert M, Lee HL, Asslander J, Zaitsev M and Hennig J 2012 Single shot concentric shells trajectories for ultra fast fMRI Magn Reson Med 68 484–94 [PubMed: 22131236]

Figure 1:

Examples of inner and outer shells of the fully sampled shells trajectory (a and b), and that of the asymmetric partial Fourier shells trajectory (c and d). The inner shells in a partial Fourier shells trajectory are the same as those in the fully sampled shells (a vs. c), while the outer shells are asymmetrically sampled along the k_z axis to achieve partial Fourier acquisition (b vs. d).

Figure 2:

Examples of shells readout from a fully sampled (FS) shells trajectory with a readout length (RO) of 512 points (a); b and c: partial Fourier (PF) trajectory design with the same readout length as the fully sampled shells (RO = 512), and a shorter readout length (RO = 384), respectively. d: 2D projection of the 3D helical spiral readouts highlighted in a, b and c onto k_y - k_z plane. The PF shells trajectories start at the same k_z as the fully sampled shells, but terminate earlier along the k_z axis. The reduced k_z sampling region can be used to increase the curvature of the trajectory (purple) which reduces the total number of interleaves required to sample the shells surface, or to reduce the number of readout points (red).

Figure 3:

a: k-Space trajectory of a partial Fourier shells corresponds to the portion that is physically sampled during data acquisition. b: the k-Space trajectory of a partial Fourier shells that is not sampled during data acquisition, but is implicitly assumed as part of the trajectory during non-iterative homodyne image reconstruction. Note that the unsampled portion in (b) shares conjugate symmetry with the corresponding part in the sampled portion (a). c: The full k-space trajectory (combining a and b) used in the proposed non-iterative reconstruction. The red circle marks the interface between the sampled and unsampled portions.

Figure 4:

ACR phantom images (3D axial acquisition reformatted into sagittal plane) acquired using a fully sampled shells trajectory (a), partial Fourier shells trajectories with an acceleration factor of 0.7 and 0.6 (b and c, respectively), reconstructed using the non-iterative homodyne reconstruction. d and e: The partial Fourier shells images reconstructed from the same k-space data as b and c after zero-filling the missing k-space samples. The line profiles in different cases are included.

Figure 5:
ACR phantom images reconstructed before and after the time-segmented NUFFT based off-resonance correction (ORC).

Figure 6:

ACR phantom images (3D axial acquisition reformatted into axial plane) acquired using fully sampled shells trajectory (a) and partial Fourier shells (acceleration factor = 0.6) with a readout length of 512 (b) and 384 (c). The resolution inserts corresponding to 1mm resolution (red arrows) are resolved in all cases. Partial Fourier shells with a shorter readout reduces the off-resonance induced image blurring (green arrows).

Figure 7:
Point spread functions generated from numerical simulation using partial Fourier accelerated shells of different readout length (512/384 points) with off-resonance (OR).

Figure 8:
Examples of in vivo MP-RAGE (a) and MP-shells images (and zoom-in insets) acquired using fully sampled shells trajectory (b) and partial Fourier accelerated shells trajectories with readout length of 512 (c) and 384 (d). The regions of interest used for white-to-gray matter contrast-to-noise ratios measurement are marked as circles.

Figure 9:

Coronal reformats of in vivo MP-RAGE (a) and MP-shells images acquired using fully sampled (FS) shells trajectory (b) and partial Fourier (PF) accelerated shells trajectories with readout length of 512 (c) and 384 (d). The PF-accelerated MP-shells with shorter readout (d) shows reduced susceptibility effect around the base of brain (arrows) compared to MP-shells with longer readout (b, c).

Table 1:

The angular rotational speed $\omega_n(\kappa)$ values for different partial Fourier acceleration factor (κ). Values calculated based on the 100th k-space shell (i.e., $n=100$) assuming the SPGR-shells acquisition setup shown in Table 2 (Readout = 512).

κ	0.6	0.7	0.8	0.9	1.0
$\omega_n(\kappa)$ (rad)	10.12	8.26	6.84	5.46	4.87

Table 2:

Details of the acquisition setup.

Sequence	Object	PF Factor	FOV (cm)	Res (mm)	BW (kHz)	FA	# of Samples per RO	ROTime (ms)	# of RO	TR/TI for MP (sec)	# of RO per TR	# of TR	Acq
SPGR-shells	ACR Phantom	Fully-Sampled	24	1	±125	8°	512	2.0	20095	-	-	-	
SPGR-shells	ACR Phantom	0.7	24	1	±125	8°	512	2.0	14111	-	-	-	
SPGR-shells	ACR Phantom	0.6	24	1	±125	8°	384	1.5	11591	-	-	-	
SPGR-shells	ACR Phantom	0.7	24	1	±125	8°	512	2.0	20253	-	-	-	
MP-RAGE	Volunteer brain	Fully-Sampled	24	1	±62.5	8°	240	2.0	240×240	2.3/0.9	240	240	
MP-shells	Volunteer brain	Fully-Sampled	24	1	±62.5	8°	512	4.1	19550	2.3/0.9	115	170	
MP-shells	Volunteer brain	0.6	24	1	±62.5	8°	512	4.1	11500	2.3/0.9	115	100	
MP-shells	Volunteer brain	0.6	24	1	±62.5	8°	384	3.1	16100	2.3/0.9	115	140	

Note: PF Factor= partial Fourier Factor; FOV = Field-of-View; Res = isotropic imaging resolution; FA = flip angle; ACR = American College of Radiology; # of RO per TR = number readout points per repetition time; # of RO is the total number of readouts for shells acquisition, or the number of phase encoding lines for MP-RAGE.

Table 3:

Image noise measurements for fully sampled (FS) MP-Shells, and partial Fourier (PF) MP-Shells with different readout length (RO) of 512 and 384. ROIs used for noise measurement are indicated as blue circles in the background regions in Figure 8.

	MPRAGE	MP-Shells (FS) RO = 512	MP-Shells (PF) RO = 512	MP-Shells (PF) RO = 384
Noise Measurement	0.082	0.077	0.095	0.095