Published in final edited form as:

NMR Biomed. 2016 May ; 29(5): 576–587. doi:10.1002/nbm.3494.

A radial sampling strategy for uniform k -space coverage with retrospective respiratory gating in 3D ultrashort-echo-time lung imaging

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Abstract

The purpose of this work was to develop a 3D radial-sampling strategy which maintains uniform k -space sample density after retrospective respiratory gating, and demonstrate its feasibility in free-breathing ultrashort-echo-time lung MRI. A multi-shot, interleaved 3D radial sampling function was designed by segmenting a single-shot trajectory of projection views such that each interleaved samples k -space in an incoherent fashion. An optimal segmentation factor for the interleaved acquisition was derived based on an approximate model of respiratory patterns such that radial interleaves are evenly accepted during the retrospective gating. The optimality of the proposed sampling scheme was tested by numerical simulations and phantom experiments using human respiratory waveforms. Retrospectively, respiratory-gated, free-breathing lung MRI with the proposed sampling strategy was performed in healthy subjects. The simulation yielded the most uniform k -space sample density with the optimal segmentation factor, as evidenced by the smallest standard deviation of the number of neighboring samples as well as minimal side-lobe energy in the point spread function. The optimality of the proposed scheme was also confirmed by minimal image artifacts in phantom images. Human lung images showed that the proposed sampling scheme significantly reduced streak and ring artifacts compared with the conventional retrospective respiratory gating while suppressing motion-related blurring compared with full sampling without respiratory gating. In conclusion, the proposed 3D radial-sampling scheme can effectively suppress the image artifacts due to non-uniform k -space sample density in retrospectively respiratory-gated lung MRI by uniformly distributing gated radial views across the k -space.

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Keywords

3D radial UTE imaging; lung MRI; retrospective respiratory gating; k -space sample uniformity

INTRODUCTION

Although radial acquisition was the first developed acquisition strategy in MRI (1,2), it has not been widely used due to its sensitivity to B_0 inhomogeneity and gradient nonlinearity, which were not tolerable in early MR systems. Thanks to the advance of MR technology, radial acquisition was recently reinstated with advantages of short echo-time (T_E) due to the absence of phase-encoding steps and low sensitivity to motion due to the oversampling around the k -space center. The absence of phase-encoding enabled the acquisition of an FID with center-out trajectory, which made radial MRI the method of choice for ultra-short T_E (UTE) imaging in diverse applications (3–8). In particular, lung imaging was the first application of radial UTE MRI, since the lung involves low proton densities, large susceptibility differences at the air–tissue interface, and respiratory motion (9–11). Radial UTE MRI improves the visualization of pulmonary tissues with a paucity of protons by minimizing the signal loss caused by T_2 decay as well as intravoxel dephasing (T_2^*) in the presence of large field inhomogeneity.

Three-dimensional lung MRI is typically performed during free breathing due to the long scan time imposed by a large field of view (FOV) and high spatial resolution. Respiratory-motion effect is a critical issue and can be suppressed via prospective or retrospective gating as in other MRI methods in the chest and abdomen (12–14). The most popular protocol of prospective gating employs a pencil-beam navigator placed on the diaphragm and accepts only the k -space segments acquired within a pre-defined displacement window while discarding and reacquiring the segments outside the acceptance window. Assuming the acquisition window is set sufficiently short, this approach ensures that all the k -space samples are acquired at the same respiratory phase (e.g. end-expiration), eliminating the risk of aliasing artifacts during the acquisition, yet results in long and varying total scan time depending on the size of the acceptance window. In contrast, retrospective gating has the advantages of relatively short and constant scan time as well as greater flexibility in choosing respiratory phases to be reconstructed, but may yield aliasing artifacts due to k -space holes created by the gating. Although radial trajectories are well suited for retrospective gating by limiting the energy of aliased signals due to the oversampling of inner k -space regions, residual artifacts of high frequency (typically with streak shape) may not be small enough to obtain high-quality images.

In retrospectively gated imaging, a k -space sampling schedule is important as it determines the location of partial k -space samples used for image reconstruction and therefore the pattern of potential aliasing artifacts. Ideally, the final sample locations should be distributed uniformly across 3D k -space so that the aliasing artifacts appear in a diffuse and benign form rather than a coherent form. A number of strategies for view scheduling have been proposed to minimize image artifacts in the context of fully sampled dynamic imaging, including bit-reversed ordering (15,16), lattice-permuted ordering (17), and golden-angle

ordering (18), but few studies have been performed in the scenario of retrospective respiratory gating. A group of retrospective approaches have been proposed, which increment the sample location in both azimuthal and polar directions by golden angles (19,20). These approaches can obtain relatively uniform sample density without any prior information, but would leave room for further improvement through subject-dependent optimization. A recent study proposed a quasi-random scheduling using 2D random numbers for the azimuthal and polar angles (21,22). However, the selection of the coefficients for random number generation was heuristic without consideration of respiratory patterns, which raises the concern of inconsistent performance over subjects.

In this study, we propose a 3D radial sampling strategy for maintaining uniform k -space coverage after retrospective respiratory gating by going through two complementary steps. First, a multi-shot, interleaved sampling function is derived in a way that uniformly distributes k -space samples within each interleaf for a given number of views per interleaf. Second, an optimal number of views per interleaf is derived based on an approximate model of a respiratory pattern such that the retrospective acceptance of k -space data occurs evenly across the obtained interleaves. The optimality of the proposed scheduling scheme is validated through numerical simulations and phantom experiments. The feasibility of free-breathing UTE lung imaging using the proposed scheme is then demonstrated in healthy volunteers.

MATERIALS AND METHODS

Sampling function for interleaved 3D radial acquisition

3D radial k -space trajectories have been designed by choosing polar and azimuthal angles in spherical coordinates in a way that uniformly covers the spherical k -space surface (23–27). The design proposed by Wong and Roos (23) has been the most widely used; this, in the scenario of interleaved acquisition, represents the normalized readout gradient waveforms as

$$\begin{aligned} G_z(p) &= \frac{2p - p_{\max} - 1}{p_{\max}}, \\ G_x(p, i) &= \cos \left(\sqrt{\frac{p_{\max}\pi}{i_{\max}}} \sin^{-1}(G_z(p)) + \frac{2i\pi}{i_{\max}} \right) \sqrt{1 - (G_z(p))^2}, \\ G_y(p, i) &= \sin \left(\sqrt{\frac{p_{\max}\pi}{i_{\max}}} \sin^{-1}(G_z(p)) + \frac{2i\pi}{i_{\max}} \right) \sqrt{1 - (G_z(p))^2} \end{aligned} \quad (1)$$

where $i_{\max}$ is the number of interleaves and $p_{\max}$ is the number of views per interleaf; p is a view index ($p = 1, 2, \dots, p_{\max}$) and i is an interleaf index ($i = 1, 2, \dots, i_{\max}$). Equations [1] provide a nearly uniform sample (or view) distribution on the k -space spherical surface when a single interleaf is used (i.e., $i_{\max} = 1$). However, as $i_{\max}$ increases from 1, a non-uniform distribution begins to appear, especially along the polar direction, because the distance between two adjacent points along the z -direction linearly increases with $i_{\max}$, as represented by black dots in Figs. 1A, C. Moreover, in the case of retrospective gating, the sample uniformity will be further degraded whenever a subset of k -space samples is discarded through the respiratory gating. The sample locations of each interleaf determine potential holes in k -space when discarded through the retrospective gating and should ideally be distributed as incoherently as possible. However, each interleaf defined in

Equations [1] traverses the spherical surface along a coherent path as represented by red and blue dots in Figs. 1A, C for the first and second interleaves, respectively, each of which consists of 30 views.

To reduce the coherence of samples within an interleaf, we design a single-shot 3D radial trajectory and segment it into $i_{\max}$ interleaves while maximizing the spacing of adjacent samples in each interleaf. Specifically, the p th point of the i th interleaf is obtained by setting $i_{\max}$ to 1 and replacing p with $(p-1)i_{\max} + i$ in Equations [1].

$$\begin{aligned} G_z(p, i) &= \frac{2((p-1)i_{\max} + i) - N_{\text{views}} - 1}{N_{\text{views}}}, \\ G_x(p) &= \cos(\sqrt{N_{\text{views}}\pi} \cdot \sin^{-1}(G_z(p))) \cdot \sqrt{1 - (G_z(p))^2}, \\ G_y(p) &= \sin(\sqrt{N_{\text{views}}\pi} \cdot \sin^{-1}(G_z(p))) \cdot \sqrt{1 - (G_z(p))^2} \end{aligned} \quad [2]$$

Figure 1B, D shows the first two interleaves (colored in red and blue) overlaid on the entire samples defined as in Equations [2]. Each interleaf distributes samples incoherently and therefore will create spatially incoherent holes in the k -space when discarded by respiratory gating.

Optimal number of views per interleaf

While the sampling functions in Equations [2] ensure uniform sample distribution within each interleaf, sample distribution across the entirety of the interleaves remains to be investigated in consideration of respiratory gating. As illustrated by the first two interleaves in Figs. 1B, D, adjacent interleaves are highly correlated in their sample locations and would yield coherent k -space holes undesirably when multiple consecutive interleaves are discarded during retrospective gating. In this section, therefore, we aim to allow the retrospective gating to accept and reject interleaves in a regular fashion by deriving an optimal number of views per interleaf as a function of respiratory and imaging parameters.

For convenience in analysis, we assume that the respiratory signal is ideally represented by a sinusoidal function whose amplitude ranges between 1 and -1 (Fig. 2). With $T_{\text{interleaf}}$ and T_{gating} defined as the duration of one interleaf and the length of the acquisition window accepted by the gating within each respiratory cycle, respectively, there are three possible cases for consideration: $T_{\text{gating}} > T_{\text{interleaf}}$ (Case I), $T_{\text{gating}} = T_{\text{interleaf}}$ (Case II), and $T_{\text{gating}} < T_{\text{interleaf}}$ (Case III). Figure 2 illustrates the retrospective selection and rejection of interleaves in each of the three cases for gating efficiencies of 50% (top row) and 33.3% (bottom row). Each interleaf is assumed to be picked when the accepted acquisition window overlaps with the center of the interleaf, as represented by black (accepted) and white (discarded) circles. While Case II ($T_{\text{gating}} = T_{\text{interleaf}}$) is most likely to yield the uniform distribution of interleaves by selecting every other interleaf (gating efficiency of 50%) or every third interleaf (gating efficiency of 33.3%), the distribution of interleaves becomes less uniform as $T_{\text{interleaf}}$ becomes shorter (Case I) or longer (Case III) than T_{gating} .

Since $T_{\text{interleaf}} = T_R p_{\max}$, the optimal $p_{\max}$ ($p_{\max, \text{opt}}$) corresponding to Case II is given by

$$p_{\max,\text{opt}} = \frac{T_{\text{gating}}}{T_{\text{R}}} \quad [3]$$

The corresponding optimal number of interleaves ($i_{\max,\text{opt}}$) is then expressed as

$$i_{\max,\text{opt}} = \frac{N_{\text{views}} T_{\text{R}}}{T_{\text{gating}}} \quad [4]$$

because $N_{\text{views}} = p_{\max} i_{\max}$. It should be noted that T_{gating} may vary during the scan, while a single representative value is needed to use Equations [3, 4]. Although the average of T_{gating} ($\langle T_{\text{gating}} \rangle$) over the entire actual scan would be a good candidate, this is not practical since the sampling scheme should be determined prior to the scan. A possible practical solution is to run a brief recording of respiratory motion ahead of the actual MR imaging and use it to determine the value of $\langle T_{\text{gating}} \rangle$.

Figure 3 illustrates respiratory-gated sample distributions (gating efficiency of 49%) and corresponding point spread functions (PSFs), obtained using the proposed sampling scheme with different numbers of views per interleaf ($p_{\max}$) (Fig. 3A–F) as well as the golden mean approach (Fig. 3G, H). With $\langle T_{\text{gating}} \rangle$ of 1890 ms and T_{R} of 3 ms, $p_{\max,\text{opt}}$ was calculated as 640. When single-spiral acquisition is used ($p_{\max} = N_{\text{views}}$ or $i_{\max} = 1$), the sample density is apparently non-uniform (Fig. 3E) and the corresponding PSF contains high side-lobe signals in the form of ridges (Fig. 3F). With the optimal number of views per interleaf ($p_{\max,\text{opt}} = 640$), the sample uniformity is substantially improved (Fig. 3A) and the corresponding PSF involves only marginal side-lobe signals (Fig. 3B). With a smaller $p_{\max}$ than the optimal number, the sample uniformity is degraded with increased side-lobe energy (Fig. 3C, D). The golden mean sampling yields a more uniform sample density than the two suboptimal cases of the proposed approach, but is inferior to the optimal case, as indicated by the wider main lobe (red arrow) and residual side lobes (yellow arrow) in the corresponding PSFs.

Numerical simulations

To verify the optimality of $p_{\max,\text{opt}}$, the sample uniformity of respiratory-gated radial data was evaluated as a function of $p_{\max}$. With N_{view} set to 157,500, $p_{\max}$ was varied from 130 to 1,180 by an increment of 50 while $i_{\max}$ was varied from 1,212 to 133 accordingly. *In vivo* respiratory waveforms were acquired for 8 min in each of six healthy subjects by acquiring a radial view along the k_z -axis periodically, as in the self-gating approach in Reference (26). For each subject, $\langle T_{\text{gating}} \rangle$ was determined by averaging the periods of gated segments and was used for calculating $p_{\max,\text{opt}}$ (Equation [3]). T_{R} was assumed to be 3 ms. For quantitative evaluation of the uniformity, we assumed an imaginary circle around a sample point on the surface of a k -space sphere, the diameter of which was 4% of the diameter of the sphere generated by Equations [2]. Then, we measured the number of points (N_{circle}) inside the circle surrounding each point. If the sample points were more uniformly distributed on the surface, N_{circle} would have a smaller variation throughout all of the points.

MRI experiments

Phantom and *in vivo* experiments were performed on a 3T whole-body scanner (Siemens Magnetom Trio, Erlangen, Germany) to demonstrate the performance of the proposed sampling strategy. The *in vivo* study was approved by the Institutional Review Board of Seoul National University Hospital, with informed consent obtained from all volunteers. A recently proposed UTE imaging sequence, a concurrent dephasing and excitation (CODE) sequence, was used for 3D radial acquisition in both phantom and *in vivo* experiments (7). CODE acquires an asymmetric gradient echo by applying the pre-dephasing gradient during RF excitation, followed by the read-out gradient. Due to the gradient-echo-based data acquisition, CODE stably provides a better image quality than other UTE techniques based on a free-induction-decay (FID) acquisition by avoiding the issue of missing a couple of sampling points at the beginning of FIDs and/or nonuniform data sampling on a gradient ramp. For all experiments, a 50 μ s long sinc pulse was used for spin excitation. The RF bandwidth of 128 kHz and gradient strength of 7.8 mT/m yielded the slab thickness of 384 mm. Image reconstruction was performed offline in MATLAB (MathWorks, R2011a, Natick, MA, USA) using gridding with the Kaiser–Bessel convolution kernel.

Phantom imaging—An ACR phantom was scanned using a four-channel head coil to evaluate the effect of k -space non-uniformity on image quality while excluding the effect of motion. While the CODE sequence was used as a primary UTE imaging method, a more traditional FID-based UTE sequence was also tested with the proposed sampling scheme. The respiratory waveform acquired for the human lung imaging was used for gating the radial data and calculating the average gating period ($\langle T_{\text{gating}} \rangle$). 89,830 views were selected out of a total number of views of 160,200 ($= N_{\text{views}}$). Two sampling cases were compared using $p_{\text{max}} = N_{\text{views}}$ ($i_{\text{max}} = 1$) and $p_{\text{max}} = p_{\text{max,opt}} = 900$ ($i_{\text{max,opt}} = 178$). Other imaging parameters were $T_E = 0.20/0.05$ ms for CODE/UTE, $T_R = 3.0$ ms, $\text{FOV} = 250 \times 250 \times 250$ mm³, flip angle = 5°, bandwidth per pixel = 530 Hz, and spatial resolution = $0.50 \times 0.50 \times 0.50$ mm³.

Human lung imaging—*In-vivo* human lung imaging was performed on nine healthy volunteers during free breathing. A three-channel body coil and a two-channel chest coil were simultaneously used for signal reception. Respiratory motion was traced by self-gating, i.e. acquiring a radial view along the k_z -axis every 30 views and thereby obtaining a 1D profile via Fourier transform in the superior–inferior direction (24). A section of the respiratory signals was acquired for 1 min prior to the main experiment for each subject. Scan parameters were $T_E/T_R = 0.14/3.0$ ms, $\text{FOV} = 384 \times 384 \times 384$ mm³, flip angle = 5°, bandwidth per pixel = 530 Hz, spatial resolution = $0.77 \times 0.77 \times 0.77$ mm³, and total scan time = 8 min. In nine subjects, end-expiratory gating was performed based on single-shot view scheduling ($p_{\text{max}} = N_{\text{views}}$) and optimized scheduling ($p_{\text{max}} = p_{\text{max,opt}}$). Three sets of images were reconstructed for each subject using the two instances of gated data as well as the ungated, fully sampled data. The three image sets were evaluated by two radiologists after they were anonymized and randomized before being presented to the reviewers. The images were scored in terms of noise, streak, and blurring based on a five-point scale. In one of the nine subjects, both end-inspiratory and end-expiratory periods were gated to show the flexibility of the retrospective gating in selecting respiratory phases to be reconstructed.

RESULTS

Numerical simulations

Figure 4A shows a representative standard deviation (STD) of N_{circle} measured in radial data gated by a subject's respiratory signal. The STD is the lowest in the sampling schedule determined by the optimal $p_{\text{max}} = p_{\text{max,opt}} = 630$, indicating the maximal k -space sample uniformity, and increases as p_{max} deviates from $p_{\text{max,opt}}$. The optimality of $p_{\text{max,opt}}$ is also demonstrated by a more concentrated histogram of N_{circle} than those obtained from the smallest and largest p_{max} values tested in the simulation (Fig. 4B). Figure 4A also shows that the STD of N_{circle} for the golden mean sampling (horizontal solid line) is lower than those for the two extreme cases of the proposed approach, but larger than the optimal case, which is consistent with the visual observation of their PSFs in Fig. 3. Figure 5 summarizes the effect of the length of the gated acquisition (T_{gating}) on the sample uniformity over six subjects' respiratory waveforms. The uniformity tends to decrease as the mean of T_{gating} decreases and the STD of T_{gating} increases, but only marginally. The weak dependence on the statistics of T_{gating} shows that the proposed sampling scheme can maintain the sample uniformity over a range of variation in breathing patterns expected in actual *in vivo* scans.

Experiments

Phantom imaging—Figure 6 shows selected axial slices of the ACR phantom obtained using UTE imaging with $p_{\text{max}} = p_{\text{max,opt}} = 900$ (A, B) and $p_{\text{max}} = N_{\text{views}} = 160,200$ (C, D), and CODE imaging with $p_{\text{max}} = p_{\text{max,opt}} = 900$ (E, F) and $p_{\text{max}} = N_{\text{views}} = 160,200$ (G, H). Figures 6B, 6D, 6F and 6H correspond to imaging slices outside the phantom and thus contain only image artifacts excluding the phantom structure. Ring- and stripe-shaped image artifacts appear in the images obtained with single-spiral acquisition ($p_{\text{max}} = N_{\text{views}}$) in both UTE and CODE sequences due to the non-uniform sample density in k -space and are well suppressed in the images obtained with the optimal views per interleaf ($p_{\text{max,opt}}$) due to the improved sample uniformity (arrows).

Human lung imaging—Figure 7 shows representative coronal (A, B, C), axial (D, E, F), and sagittal (G, H, I) images of a healthy human lung (Volunteer 1) obtained using full data without respiratory gating (A, D, G), and gated data with $p_{\text{max}} = N_{\text{views}}$ (B, E, H) and $p_{\text{max}} = p_{\text{max,opt}} = 900$ (C, F, I). Compared with the full-data reconstruction, the proposed method significantly improves the depiction of segmental bronchial walls (Arrows 2 and 9) as well as segmental and subsegmental pulmonary vessels (Arrows 3, 6, and 8) due to the reduction of motion-related blurring through respiratory gating. Despite k -space undersampling by a factor of 0.56 (a ratio of selected views to total views), streak artifacts are barely seen due to the diffused aliasing pattern yielded by the uniform sample density. Compared with respiratory-gated single-spiral acquisition, which involves the same undersampling factor, the optimized sampling scheme significantly reduces the streak (Arrows 1 and 7) and ringing artifacts (Arrow 5), improving overall signal-to-noise ratio and making the boundary of the diaphragmatic dome clearer (Arrow 4).

Figure 8A shows the quantitative sample uniformity measured in all nine subjects for the two cases of $p_{\text{max}} = N_{\text{views}}$ (circles) and $p_{\text{max}} = p_{\text{max,opt}}$ (triangles). The STD of N_{circle} is

lower with the optimal $p_{\max}$ consistently over all subjects. Figure 8B summarizes the qualitative analysis of the *in vivo* images obtained with full sampling (light grey) and proposed gating with $p_{\max} = N_{\text{views}}$ (dark grey) and $p_{\max} = p_{\max, \text{opt}}$ (black). The proposed retrospective gating with the optimal number of views per interleaf obtained higher scores than the gating with single-spiral sampling and full sampling in all the metrics (noise, streak and blurring).

Figure 9 shows coronal lung images of another healthy subject (Volunteer 2) reconstructed in the end-inspiratory (Fig. 9A) and end-expiratory (Fig. 9B) phases of the lung in motion. Figure 9C illustrates the retrospective gating windows for the two phases on the respiratory waveforms extracted from the self-gating. Respiration-induced expansion (Fig. 9A) and contraction (Fig. 9B) of the lung are clearly visualized with only marginal streak artifacts. For reference, the displacement of the diaphragm is indicated by two dashed lines for the right dome of the diaphragm in inspiration (yellow) and expiration (red).

DISCUSSION

The proposed sampling strategy aimed to incoherently distribute potential k -space holes created by retrospective respiratory gating so as to yield PSF side-lobes with small and diffused magnitude and therefore minimal aliasing artifacts. The incoherence across the entire k -space was achieved by pursuing incoherence in sample distribution within each interleaf as well as across neighboring interleaves. The intra-interleaf incoherence was maintained through multi-interleaved radial trajectories designed by segmenting a single-spiral trajectory in an appropriate manner. The inter-interleaf coherence was shown to be minimized by optimizing the segmentation factor for the interleaved acquisition based on respiratory and gating parameters. That is, the duration of one interleaf ($T_R p_{\max}$) should be equal to the length of the accepted acquisition window in each respiratory cycle.

Breathing motion is expected to be diverse in pattern across subjects, especially in patients with dyspnea. From the perspective of gated sampling, the pattern of respiratory motion can be characterized by the average and variation of T_{gating} over time. The duration of T_{gating} , only if kept unchanged over time, would not significantly degrade the uniformity of sampled k -space positions, as explained in Fig. 2 and related text. Rather, it determines an undersampling factor, affecting the energy of aliased signals. On the other hand, variation of T_{gating} during the scan may significantly affect the performance of the proposed sampling method by altering the optimal condition over time. In Fig. 5, the numerical simulation based on respiratory waveforms of six subjects showed the trend that the degree of non-uniformity measured by the STD of N_{circle} (the number of nearby samples) only slightly increases in proportion to the STD of T_{gating} .

A slightly cumbersome requirement of the proposed method is the acquisition of a section of the respiratory signals ahead of running the main sequence to determine the averaged gating period ($\langle T_{\text{gating}} \rangle$). While the additional time for the scout measurement and processing is minor (e.g. 1 min), inaccurate estimates of $\langle T_{\text{gating}} \rangle$ may degrade the uniformity of the k -space samples scheduled by the proposed scheme. The effect of $\langle T_{\text{gating}} \rangle$ difference on sample uniformity can be inferred by examining the plot of $p_{\max}$ versus STD of N_{circle} (Fig.

4A). Apparently, inaccurate estimation of $\langle T_{\text{gating}} \rangle$ will increase the deviation of p_{max} from $p_{\text{max,opt}}$ and therefore the STD of N_{circle} , implying a decrease in sample uniformity. However, it should be noted that the STD of N_{circle} remains small in a large neighborhood around the optimal condition. Collectively, over the six subjects' respiratory waveforms tested in this study, we found that a $\langle T_{\text{gating}} \rangle$ -estimation error of ± 300 ms changes the STD of N_{circle} only by $12.18 \pm 6.82\%$ (or 0.16 ± 0.08 in absolute numbers), which supports the robustness of the proposed method with regard to the potential error in the scout measurement. One possible approach to remove the need for a prescan is real-time scheduling of the radial views, as in previous studies (28,29). Once the self-gating signal used in this study is reconstructed on the scanner with minimal latency, the optimal condition for uniform sample density can be updated on the fly. Another possibility for needing no prescan might be available from a more practical standpoint: one can use an average of previously known breathing patterns of healthy or diseased people without running a prescan, since the proposed method is robust in a large neighborhood of the optimal condition (Fig. 4A).

The proposed sampling strategy has potential for applications other than the lung imaging shown in the present study. Imaging of any organs that are susceptible to respiratory motion can benefit from retrospective respiratory gating with the proposed sampling scheme. Quantitative analysis of dynamic contrast-enhanced MRI in the lung and upper abdominal organs may be achieved more robustly by the proposed method without loss of temporal resolution or motion artifacts (30). Sequential and separated reconstruction of acquired MR data according to the respiratory phase may allow for fully 3D isotropic volumetric evaluation of dynamic change of chest wall, diaphragm, lung volume (31,32), and lung cancer limitedly assessed by current MR sequences (33). 3D cardiac imaging appears to be another promising application, since it suffers from excessively low scan efficiency due to the need for both cardiac and respiratory gatings (34). Specific potential applications include 3D late gadolinium enhancement imaging and contrast-enhanced coronary artery imaging. The proposed sampling scheme can also be combined with k -space trajectories other than radial readouts. The key requirement for eligibility is to oversample the inner k -space so as to yield aliasing artifacts with low energy and well-diffused pattern when uniformly undersampled using the proposed scheme. In general, 3D center-out trajectories are well suited for this requirement, including 3D cone trajectories and generalized 3D spiral trajectories (35–37). Last, although our sampling strategy was based on the 3D radial trajectory proposed by Wong and Roos, it can also be applied to other interleaved sampling functions, including golden-angle-ordered readout in a spiral phyllotaxis pattern (27).

Prospective gating has the advantage that full sampling of k -space is always guaranteed, eliminating the issue of aliasing artifacts, but also involves issues of prolonged scan times and/or respiratory motion artifacts depending on the length of the gated acquisition window. If the acquisition window is sufficiently short, motion artifacts can be avoided, albeit at the cost of increased scan time. If the acquisition window is widened to mitigate the scan time penalty, the risk of respiratory motion artifacts will be increased as a trade-off. While the clinical utility of the proposed method needs to be further validated through comparisons with the standard prospective gating, it would be necessary to investigate the effect of acquisition length on resulting motion artifact and scan time.

In conclusion, a 3D radial-sampling scheme was proposed to maintain uniform k -space sample density in retrospectively respiratory-gated imaging and thus minimize aliasing artifacts. This was achieved by using a single-spiral-segmented sampling function with the optimized segmentation factor based on respiratory and gating parameters. Phantom and human lung experiments demonstrated that the proposed sampling scheme suppresses streak and ringing artifacts well while reducing motion-related blurring. While the proposed scheme was demonstrated in 3D radial lung MRI here, it has the potential for other applications, such as free-breathing cardiac MRI and other center-out k -space acquisitions.

Acknowledgments

This work was supported by NRF Grants 2010-0025744, IBS-R015-D1, and NIH R21 EB019206.

Abbreviations used

T_E	echo time
UTE	ultra-short T _E
FOV	field of view
T_R	repetition time
PSF	point spread function
CODE	concurrent dephasing and excitation
FID	free induction decay
STD	standard deviation

References

1. Lauterbur PC. Image formation by induced local interactions: examples employing nuclear magnetic resonance. *Nature*. 1973; 242:190–191.
2. Lai CM, Lauterbur PC. True three-dimensional image reconstruction by nuclear magnetic resonance zeugmatography. *Phys Med Biol*. 1981; 26(5):851–856. [PubMed: 7291305]
3. Pauly, JM.; Conolly, SM.; Nishimura, DG.; Macovski, A. Slice-selective excitation for very short T2 species. Proceedings of ISMRM 8th Annual Meeting; Amsterdam, Netherlands. 1989; p. 28
4. Glover GH, Pauly JM, Bradshaw KM. Boron-11 imaging with a three-dimensional reconstruction method. *J Magn Reson Imaging*. 1992; 2(1):47–52. [PubMed: 1623280]
5. Idiyatullin D, Corum C, Park JY, Garwood M. Fast and quiet MRI using a swept radiofrequency. *J Magn Reson*. 2006; 181(2):342–349. [PubMed: 16782371]
6. Wu Y, Dai G, Ackerman JL, Hrovat MI, Glimcher MJ, Snyder BD, Nazarian A, Chesler DA. Water- and fat-suppressed proton projection MRI (WASPI) of rat femur bone. *Magn Reson Med*. 2007; 57(3):554–567. [PubMed: 17326184]
7. Park JY, Moeller S, Goerke U, Auerbach E, Chamberlain R, Ellermann J, Garwood M. Short echo-time 3D radial gradient-echo MRI using concurrent dephasing and excitation. *Magn Reson Med*. 2012; 67(2):428–436. [PubMed: 21702064]
8. Weiger M, Wu M, Wurnig MC, Kenkel D, Jungraithmayr W, Boss A, Pruessmann KP. Rapid and robust pulmonary proton ZTE imaging in the mouse. *NMR Biomed*. 2014; 27(9):1129–1134. [PubMed: 25066371]
9. Mayo JR, MacKay A, Müller NL. MR imaging of the lungs: value of short TE spin-echo pulse sequences. *Am J Roentgenol*. 1992; 159(5):951–956. [PubMed: 1414805]

10. Bergin CJ, Pauly JM, Macovski A. Lung parenchyma: projection reconstruction MR imaging. *Radiology*. 1991; 179(3):777–781. [PubMed: 2027991]
11. Johnson KM, Fain SB, Schiebler ML, Nagle S. Optimized 3D ultrashort echo time pulmonary MRI. *Magn Reson Med*. 2013; 70(5):1241–50. [PubMed: 23213020]
12. Ehman RL, McNamara MT, Pallack M, Hricak H, Higgins CB. Magnetic resonance imaging with respiratory gating: techniques and advantages. *Am J Roentgenol*. 1984; 143(6):1175–1182. [PubMed: 6333787]
13. McConnell MV, Khasgiwala VC, Savord BJ, Chen MH, Chuang ML, Edelman RR, Manning WJ. Comparison of respiratory suppression methods and navigator locations for MR coronary angiography. *Am J Roentgenol*. 1997; 168(5):1369–1375. [PubMed: 9129447]
14. Puderbach M, Hintze C, Ley S, Eichinger M, Kauczor HU, Biederer J. MR imaging of the chest: a practical approach at 1.5 T. *Eur J Radiol*. 2007; 64(3):345–355. [PubMed: 17900843]
15. Spielman DM, Pauly JM, Meyer CH. Magnetic resonance fluoroscopy using spirals with variable sampling densities. *Magn Reson Med*. 1995; 34(3):388–394. [PubMed: 7500878]
16. Rasche V, de Boer RW, Holz D, Proksa R. Continuous radial data acquisition for dynamic MRI. *Magn Reson Med*. 1995; 34(5):754–761. [PubMed: 8544697]
17. Tsao J, Boesiger P, Pruessmann KP. Lattice permutation for reducing motion artifacts in radial and spiral dynamic imaging. *Magn Reson Med*. 2006; 55:116–125. [PubMed: 16323156]
18. Winkelmann S, Schaeffter T, Koehler T, Eggers H, Doessel O. An optimal radial profile order based on the Golden Ratio for time-resolved MRI. *IEEE Trans Med Imaging*. 2007; 26:68–76. [PubMed: 17243585]
19. Rahmer, J.; Keupp, J.; Caruthers, SD. Self-navigated motion compensation in simultaneous 19F/1H 3D radial imaging using golden means profile interleaving. *Proceedings of ISMRM 16th Annual Meeting*; Toronto, Canada. 2008; p. 1471
20. Chan RW, Ramsay EA, Cunningham CH, Plewes DB. Temporal stability of adaptive 3D radial MRI using multidimensional golden means. *Magn Reson Med*. 2009; 61:354–363. [PubMed: 19165897]
21. Weick, S.; Ehses, P.; Blaimer, M.; Brueur, FA.; Jakob, PM. Improved retrospective self-gated human lung imaging using a quasi random sampling scheme. *Proceedings of ISMRM 19th Annual Meeting*; Toronto, Canada. 2011; p. 924
22. Hemberger, K.; Weick, S.; Breuer, FA.; Jakob, PM.; Haddad, D. Retrospective self-gated 3D UTE lung imaging. *Proceedings of ISMRM 21th Annual Meeting*; Salt Lake City, UT, USA. 2013; p. 1489
23. Wong ST, Roos MS. A strategy for sampling on a sphere applied to 3D selective RF pulse design. *Magn Reson Med*. 1994; 32(6):778–84. [PubMed: 7869901]
24. Barger AV, Block WF, Toropov Y, Grist TM, Mistretta CA. Time-resolved contrast-enhanced imaging with isotropic resolution and broad coverage using an undersampled 3D projection trajectory. *Magn Reson Med*. 2002; 48(2):297–305. [PubMed: 12210938]
25. Stehning C, Börnert P, Nehrke K, Eggers H, Dössel O. Fast isotropic volumetric coronary MR angiography using free-breathing 3D radial balanced FFE acquisition. *Magn Reson Med*. 2004; 52(1):197–203. [PubMed: 15236387]
26. Stehning C, Börnert P, Nehrke K, Eggers H, Stuber M. Free-breathing whole-heart coronary MRA with 3D radial SSFP and self-navigated image reconstruction. *Magn Reson Med*. 2005; 54(2):476–480. [PubMed: 16032682]
27. Piccini D, Littmann A, Nilles-Vallespin S, Zenge MO. Spiral phyllotaxis: the natural way to construct a 3D radial trajectory in MRI. *Magn Reson Med*. 2011; 66(4):1049–1056. [PubMed: 21469185]
28. Manduca A, McGee KP, Welch EB, Felmlee JP, Grimm RC, Ehman RL. Autocorrection in MR imaging: adaptive motion correction without navigator echoes. *Radiology*. 2000; 215:904–909. [PubMed: 10831720]
29. Stehning, C.; Dannels, W.; Bornert, P.; Nehrke, K.; Dossel, O. Free breathing MRI using isotropic 3D-radial sampling with motion-dependent readout orientation (ROAR). *Proceedings of ESMRMB*; Copenhagen, Denmark. 2004; p. 251

30. Chandarana H, Feng L, Block TK, Rosenkrantz AB, Lim RP, Babb JS, Sodickson DK, Otazo R. Free-breathing contrast-enhanced multiphase MRI of the liver using a combination of compressed sensing, parallel imaging, and golden-angle radial sampling. *Invest Radiol.* 2013; 48(1):10–16. [PubMed: 23192165]
31. Plathow C, Schoebinger M, Fink C, Ley S, Puderbach M, Eichinger M, Bock M, Meinzer H, Kauczor H. Evaluation of lung volumetry using dynamic three-dimensional magnetic resonance imaging. *Invest Radiol.* 2005; 40(3):173–179. [PubMed: 15714092]
32. Tokuda J, Schmitt M, Sun Y, Patz S, Tang Y, Mountford CE, Hata N, Wald LL, Hatabu H. Lung motion and volume measurement by dynamic 3D MRI using a 128-channel receiver coil. *Acad Radiol.* 2009; 16(1):22–27. [PubMed: 19064208]
33. Dinkel J, Hintze C, Tetzlaff R, Huber PE, Herfarth K, Debus J, Kauczore HU, Thieke C. 4D-MRI analysis of lung tumor motion in patients with hemidiaphragmatic paralysis. *Radiother Oncol.* 2009; 91(3):449–454. [PubMed: 19394712]
34. Biglands JD, Radjenovic A, Ridgway JP. Cardiovascular magnetic resonance physics for clinicians: part II. *J Cardiovasc Magn Reson.* 2012; 14:66. [PubMed: 22995744]
35. Gurney PT, Hargreaves BA, Nishimura DG. Design and analysis of a practical 3D cones trajectories. *Magn Reson Med.* 2006; 55:575–582. [PubMed: 16450366]
36. Pipe JG, Zwart NR, Aboussouan EA, Robison RK, Devaraj A, Johnson KO. A new design and rationale for 3D orthogonally oversampled k -space trajectories. *Magn. Reson. Med.* 2011; 66:1303–1311.
37. Turley DC, Pipe JG. Distributed spirals: a new class of three-dimensional k -space trajectories. *Magn. Reson. Med.* 2013; 70:413–419.

Figure 1.

Distribution of 3D radial samples on the k -space sphere based on the original (A, C; Equations [1]) and the proposed (B, D; Equations [2]) sampling functions displayed in an oblique view (A, B) and on the k_x - k_z plane (C, D). For clear illustration, a relatively small number of views ($N_{\text{views}} = 3000$) was used with 100 interleaves ($i_{\text{max}} = 100$). While all of the samples are represented by black dots, the first two interleaves are colored by red and blue dots. Using the proposed sampling function, each of the two interleaves distributes samples in a more incoherent fashion (B, D) and would create more incoherent k -space holes *within each interleaf* when omitted due to the respiratory gating. Note that the spatial correlation *between the two interleaves* remains high even with the proposed sampling function. It is therefore desired to accept and reject interleaves in a consistent order to maintain sample uniformity across all the accepted interleaves.

Figure 2.

Schematic diagram of retrospective respiratory gating of k -space interleaves under the assumption of sinusoidal respiratory waveforms with gating efficiency of 50% (top row) and 33.3% (bottom row). With $T_{\text{interleaf}}$ and T_{gating} defined as the duration of one interleaf and the length of the acquisition window accepted by the gating within each respiratory cycle, respectively, three cases of $T_{\text{gating}} > T_{\text{interleaf}}$, $T_{\text{gating}} = T_{\text{interleaf}}$, and $T_{\text{gating}} < T_{\text{interleaf}}$ were simulated. Each interleaf was accepted if the gating window overlapped with the center of the interleaf, as represented by a black circle; otherwise, it was discarded, as represented by a white circle. The selected interleaves are evenly distributed when $T_{\text{interleaf}}$ is equal to T_{gating} (B, E). As $T_{\text{interleaf}}$ deviates from T_{gating} , the distribution of the chosen interleaves becomes less uniform (A, C, D, F).

Figure 3. Distributions of retrospectively gated radial data using an *in vivo* respiratory signal (1st column), and corresponding PSFs (2nd column). A total of 30,000 views (N_{views}) were simulated. Using the proposed sampling scheme, the k -space sample density is the most uniform with the optimal views per interleaf ($p_{\text{max,opt}}$) (A) while becoming non-uniform when p_{max} is smaller (C) or larger (E) than $p_{\text{max,opt}}$. Accordingly, the side-lobe of the PSF is the most evenly distributed with the smallest amplitude when $p_{\text{max,opt}}$ is used (B), compared with the other two cases (D and F). Using the golden mean approach, the sample density appears to be as uniform as in the optimal case (A vs G) but yields wider mainlobe (red arrow) and larger sidelobe signals (yellow arrow) in the PSF.

Figure 4.

STD of the number of neighboring points (N_{circle}) as a measure of sample uniformity in gated radial data using the proposed sampling (solid line with circles) and golden-means-based (horizontal solid line) schemes with six subjects' respiratory waveforms. (A) The STD measured with a representative subject's respiratory signal is minimal (i.e. maximum uniformity) at $p_{\max} = p_{\max, \text{opt}}$ and increases as $p_{\max}$ deviates from $p_{\max, \text{opt}}$. The golden mean sampling achieves relatively low STD but is higher than the optimal case of the proposed approach. (B) The histograms of N_{circle} for three $p_{\max}$ values show the most concentrated distribution at $p_{\max} = p_{\max, \text{opt}}$ (red) than the two suboptimal cases (yellow and blue).

Figure 5.

Effect of the mean and variation of the gated acquisition period (T_{gating}) on sample uniformity (STD of N_{circle}) measured with six subjects' respiratory waveforms using the proposed sampling strategy. Sample uniformity tends to decrease as the mean of T_{gating} ($\langle T_{\text{gating}} \rangle$) decreases (A) and the STD of T_{gating} increases (B), but very slightly, which supports robust performance of the proposed scheme over various respiratory patterns.

Figure 6.

Phantom images reconstructed with gated data obtained using UTE (A–D) and CODE (E–H) sequences. For each sequence, $p_{\max} = p_{\max, \text{opt}}$ and $p_{\max} = N_{\text{views}}$ were used. Two slices are shown, where one is taken outside the phantom and therefore shows only the undesired signal created by sample non-uniformity (e.g. ringing and streak artifacts), excluding the phantom structure. In both sequences, the artifacts caused by sample non-uniformity were significantly reduced using the proposed sampling method with the optimal $p_{\max}$, as pointed out by white arrows.

Figure 7.

Pulmonary images in a healthy subject (Volunteer 1) reconstructed using full radial data without respiratory gating (first column), retrospectively respiratory-gated data with single-spiral sampling ($p_{\max} = N_{\text{views}}$) (second column), and the optimally interleaved sampling ($p_{\max} = p_{\max, \text{opt}}$) (third column). The retrospective gating efficiency was 56%. Compared with the full sampling, the proposed method improves the depiction of segmental bronchial walls (Arrows 2 and 9) and pulmonary vessels (Arrows 3, 6, and 8) due to reduced motion-related blurring and streak artifacts. Compared with the respiratory gating with single-spiral sampling, the proposed sampling method significantly suppresses the aliasing artifacts that appear as streaks (Arrows 1 and 7) or rings (Arrow 5), due to improved uniformity of k -space coverage.

Figure 8.

Results of quantitative sample uniformity (STD of N_{circle}) and qualitative image quality evaluation. (A) The STD of N_{circle} is lower with $p_{\max} = p_{\max, \text{opt}}$ (triangles) than $p_{\max} = N_{\text{views}}$ (circles) consistently over all the nine subjects. (B) The proposed retrospective gating with the optimal sampling (black) obtained higher scores than the retrospective gating with single-spiral sampling (dark grey) and full sampling without gating (light grey) in all qualitative metrics (noise, streak, and blurring). While the suboptimal gating received lower scores for noise and streak than the full sampling due to less k -space data used for image reconstruction, the suboptimal gating was less blurry due to reduced motion effects.

Figure 9.

(A, B) Coronal lung images of a healthy subject (Volunteer 2) reconstructed in the end-inspiration (A) and the end-expiration (B) phases with gating efficiencies of 24.7% and 25.2%, respectively. Expansion and contraction of the lung during the respiration are clearly identified in A and B, respectively. The right dome of the diaphragm in inspiration and expiration is indicated by yellow and red lines, respectively. (C) The self-gated respiratory signal is displayed along with the gating windows used for inspiratory (yellow box) and expiratory (red box) reconstructions.