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Evaluation of Partial k -space strategies to speed up Time-domain EPR Imaging

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Abstract

Narrow-line spin probes derived from the trityl radical have led to the development of fast *in vivo* time-domain EPR imaging. Pure phase-encoding imaging modalities based on the Single Point Imaging scheme (SPI) have demonstrated the feasibility of 3D oximetric images with functional information in minutes. In this paper, we explore techniques to improve the temporal resolution and circumvent the relatively short biological half-lives of trityl probes using partial k -space strategies. There are two main approaches: one involves the use of the Hermitian character of the k -space by which only part of the k -space is measured and the unmeasured part is generated using the Hermitian symmetry. This approach is limited in success by the accuracy of numerical estimate of the phase roll in the k -space that corrupts the Hermiticity. The other approach is to measure only a judiciously chosen reduced region of k -space (a centrosymmetric ellipsoid region) that more or less accounts for >70% of the k -space energy. Both of these aspects were explored in FT-EPR imaging with a doubling of scan speed demonstrated by considering ellipsoid geometry of the k -space. Partial k -space strategies help improve the temporal resolution in studying fast dynamics of functional aspects *in vivo* with infused spin probes.

Keywords

MRI; Time-domain EPR imaging; Partial Fourier; partial k -space acquisition; conjugate symmetry; Single Point Imaging; homodyne detection; projection onto convex sets (POCS)

1. Introduction

Electron Paramagnetic Resonance Imaging (EPRI) is an emerging technique for *in vivo* imaging of tissue pO_2 in experimental animals with potential for human applications. Recent

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developments in pulsed EPRI (1–5) and the advent of narrow line spin probes based on trityls (6) have improved the resolution of EPRI to sub-millimeter levels. Fourier Transform (FT)-EPRI techniques allow faster scanning by an order of magnitude compared to conventional CW techniques facilitating dynamic imaging of tissue pO_2 within minutes (7). Further improvement of scan speed is highly desirable for *in vivo* EPRI to permit studies of temporal dynamics of tumor pO_2 . Usually a limited time window is available for scanning under optimal conditions *in vivo* depending on the pharmacokinetics of the injected paramagnetic tracer (8). This poses a serious limitation on the ability to pursue dynamically changing biological phenomena. Reduction of data acquisition time is an active area of research in all imaging modalities including EPRI.

FT-EPRI using the Single Point Imaging (SPI) method directly measures k -space image data similar to MRI where partial Fourier (PF) techniques (9–13) and partial k -space acquisition and reconstruction (PAR) are well established to enhance scan speed (14–21). PF algorithms exploit the Hermitian symmetry of k -space which relates one half of the k -space to the other half by its complex conjugate (9–13). PAR algorithms either acquire the part of k -space that is sufficient to reconstruct the image (15) or use spiral trajectories (16,20). McGibney *et al.* (12) evaluated the performance of PF algorithms including conjugate synthesis, the Margosian method, homodyne detection, and projection onto convex sets (POCS). Application of spiral acquisition techniques to single point imaging (SPI) has been described by Halse *et al.* (16).

Further improvement in scan speed may be achieved for sparse images by compressed sensing techniques (22,23), reconstruction from arbitrary k -space (19) and model based image reconstruction (14), or optimization of k -space trajectory (21). These are computationally more intensive and the k -space acquisition is optimized for performance in reconstruction. Though EPRI can potentially take advantage of many of these algorithms, these are yet to be evaluated for practical EPRI applications. In this study, we applied well-known PF and PAR algorithms that are routinely used in MRI as techniques to potentially reduce image acquisition time.

Although MRI and FT-EPRI share the same fundamentals, there are significant differences between these two technologies which call for performance evaluation of PAR and PF algorithms in EPRI. EPRI deals with transverse relaxation times on the order of microseconds (3) in contrast to conventional MRI that deals with decays on the order of tens to hundreds of milliseconds. Because of these short relaxation times, SPI techniques are used to improve image quality. The k -space matrix size and resolution at the current state of art are relatively lower in EPRI compared to MRI. Additionally, the B_0 field strength in EPRI is several orders of magnitude lower than that used in MRI, resulting in different effects due to magnetic susceptibility (which is the primary cause of phase error in MRI and limits PF techniques). Due to the large sampling bandwidth and low coil Q necessary for data acquisition in EPRI, artifacts arising from shifts in signal levels and random spikes from other sources are often better controlled in MRI than EPRI. Furthermore, EPRI typically has relatively lower SNR values than proton MRI. All of these differences may affect the performance of PF and PAR algorithms in EPRI. The aim of this study is to

identify an appropriate PAR algorithm to reduce EPRI acquisition time without the need for extensive computation.

2. Methods

2.1 Phantom/animal preparation

For 2D measurements, a resolution phantom made of a Lucite cylinder with rectangular partitions having minimum slot width of 300 μm filled with 3 mM Oxo-63 solution (GE Healthcare) was used. Three dimensional *in vivo* EPRI measurements were performed on a tumor-bearing hind leg of a female C3H mouse. Details of the phantom and animal handling including tumor growth using SCCVII tumor cells to about 1.5 cm diameter in one of the hind legs, animal care, mounting in the resonator, infusion of spin probe through tail vein, maintenance of body temperature and monitoring of breathing rate have been previously described (24). In both the phantom and the mouse, full k -space data were obtained to allow comparisons between full and partial k -space reconstructions. Partial k -space scanning was performed on a five tube phantom filled with 2 mM trityl aqueous solution at different oxygen concentrations which was placed inside a specially made holder.

2.2 Imaging

The image data acquisition was performed in 2 and 3 dimensions on an FT-EPR imager operating at 300 MHz in SPI mode as previously described (2). For imaging of phantoms in 2D, gradients were applied in the transverse plane (perpendicular to the axes of the tubes) in equal increment steps covering the full range of G_{max} to $-G_{\text{max}}$, where G_{max} is the maximum gradient magnitude along each direction. For each gradient setting, 4000 FID's were acquired in quadrature and summed under CYCLOPS phase cycling (25). The resolution phantom image was scanned at matrix size of 61×61 and $G_{\text{max}} = 40$ mT/m. Baseline shifts were removed from k -space data to reduce the DC offset artifacts in the middle of the image. 3D *in vivo* imaging of a mouse tumor was performed using matrix size = $19 \times 19 \times 19$, $\tau = 900$ ns and $G_{\text{max}} = 14$ mT/m. *In vivo* oximetric imaging was performed using the same matrix size, but at three G_{max} values of 9.6, 12 and 14 mT/m. Full details of SPI measurement and oximetry have been described previously (1,24,26). For partial k -space acquisitions of the five tube phantom, 2D imaging was performed with matrix size of 21×21 , $\tau = 850$ ns and $G_{\text{max}} = 8, 10$, and 12 mT/m.

The EPR imager is versatile to scan the k -space in any desired trajectory and geometry and hence acquisition of k -space data with various trajectories including Cartesian, elliptical, spiral, or any other arbitrary k -space is relatively straightforward. Cartesian grid and elliptical k -space data were acquired by optimizing the trajectory to avoid abrupt changes in gradient magnitude or sign. Data acquisition was divided into two steps: (i) Selection of k -space points to be scanned and (ii) determination of scanning trajectory to optimize gradient changes. This two-step process facilitates flexibility in selection of k -space points and optimization of trajectory for desired experimental variables. In a 3D full k -space data collection, an FID with the Zeeman field off-set far from resonance is acquired to be used as background prior to acquiring data for successive 2D planes. In the partial k -space data collection we use similar strategy of collecting one background FID for every step of the

third gradient. The partial k -space data acquired was converted to full k -space by substituting the k -space points where data was not collected with background to achieve zero filling after background correction step. Calculations for the application of PF and PAR algorithms and image reconstruction were done using MATLAB scripts (The Mathworks Inc., Natick, MA) developed in-house as necessary.

3. Results and Discussion

3.1 Review of PF and PAR algorithms

PF and PAR algorithms to increase acquisition speed have been described, reviewed and refined over the past three decades and are still being explored for MRI applications (9–23,27). The performance of PF algorithms exploiting Hermitian symmetry depends on the accuracy of phase corrections and merging filters (9–13). On the other hand, numerous PAR algorithms are described considering non-Cartesian sampling patterns in radial geometry (20, 28–30), compressed sensing techniques for sparse images (22, 23), and reconstruction from arbitrary k -space (19). Among these, the spiral acquisition and reconstruction procedures have been extensively studied and successfully implemented for fast acquisition (15,16,20). However, we have limited the scope of this work only to a few common algorithms for routine EPRI applications. A brief discussion of PF and PAR algorithms and their performance in EPRI are given below.

3.1.1 PF algorithms—Data acquisition of half of the k -space, in principle, is redundant because of the Hermitian symmetry of k -space that relates one half of the k -space to the other half by its complex conjugate. However, phase errors arising from several sources during the measurement including frequency variation, local susceptibility changes, and motion are to be estimated and corrected to ensure the conjugate symmetry. PF algorithms were extensively studied and their limitations were documented in literature (9–13) and reviewed by McGibney *et al.*(12).

In *conjugate synthesis*, one half of the unknown k -space data are replaced by the complex conjugate of the known other half. The image reconstructed by inverse DFT of these data generally contains artifacts such as Gibbs' ringing that arises from sharp changes in k -space. *Margosian* PF algorithm and *homodyne detection* methods overcome this problem to a certain extent (12). The Margosian algorithm applies a merging filter that smoothes the transition between the acquired and synthesized data. Image reconstruction is done by inverse DFT followed by multiplication of a phase estimator function and magnitude estimates are presented as image intensities. The homodyne detection algorithm described by Noll *et al.* (13) is similar to this algorithm but uses the real part of the image data after appropriate phase correction. This algorithm provides better resolution at lower intensities than the others. Both these algorithms require phase values which are estimated by inverse DFT of a symmetric k -space region at low spatial frequencies and perform well when the image phase variations are small. When the image phase is varying rapidly, the phase correction estimates to the real part at one pixel may be affected by the imaginary part of neighboring pixels. Single step phase correction procedures perform poorly in such conditions.

Iterative methods using projection onto convex sets (POCS) can help to overcome this problem (12). In these methods, the phase of the reconstructed image is constrained to a phase estimate derived from symmetric portion while estimating the missing k -space data iteratively. In a simple version of POCS algorithm (11), the steps of inverse DFT from spatial frequency domain to image domain, phase correction using symmetric region of k -space, DFT of phase corrected image, and replacement of unknown k -space with DFT estimates are repeated for each iteration. At convergence, a merging filter is applied, image data is inverse Fourier transformed and either the real part or magnitude is presented as image.

McGibney *et al.* pointed out that phase estimates may lead to smearing due to circular convolution and preferred direct convolution in spatial frequency domain (12). This method is equivalent to passing the partial data through a finite impulse response filter that reduces the smearing. In summary, the exploitation of Hermitian symmetry in EPRI is limited by the extent we can correctly do the phase corrections and may also lead to loss of spatial resolution.

3.1.2 PAR algorithms

3.1.2.1 Optimization of the k -space: The phase estimation problem of PF algorithms can be circumvented by acquiring symmetric k -space region in a specific geometry. Most of the k -space energy is concentrated at low spatial frequencies which contribute to the intensity of the image whereas the high spatial frequency region contributes to image resolution. Interestingly, partial k -space acquisition of low frequency regions alone was shown to improve SNR in fMRI (17). The acquisition speed of this procedure is similar to that of spiral acquisition. Further, it does not require estimation of density compensation function (DCF) to adjust for sampling density variations (31) and avoids computational costs and the reconstruction artifacts resulting from regridding of spiral data to a uniform Cartesian grid.

3.1.2.2 Spiral trajectory: Data acquisition in non-uniformly spaced trajectories such as Archimedean spiral (16,20) minimizes gradient vibration allowing faster scan speed and was found to minimize motion artifacts in the image due to the subject's movement by breathing and peristalsis. The spiral trajectory begins at the origin and proceeds towards periphery by angular increments of θ and radial increments of r through revolution. Image reconstruction procedures of single point MRI in spiral trajectory were described by Balcom and others (16,20). Briefly, the procedure is: (i) weighting the data by a DCF to account for higher density of points at the origin relative to periphery (ii) re-gridding spiral data to uniformly spaced Cartesian grid by convolution of discrete spiral data with a continuous function such as Kaiser-Bessel (iii) Inverse DFT and de-apodization. The performance of this method depends on the quality of DCF, choice of the convolution function and de-apodization. The acquisition time is shorter by 21% ($\approx 1 - \pi/4$) since corners of the rectangular grid are not explored. Dense sampling of k -space near the origin where the energy is high results in better SNR. Image reconstruction of spiral data however, is computationally more intensive than acquisition in Cartesian grid. Other non-Cartesian sampling patterns such as radial, Lissajou and rosette are also described in literature (20,28–30).

3.2 Performance evaluation of PF and PAR algorithms for EPRI applications

Two-dimensional full k -space EPRI data of the phantom acquired at a matrix size of 61×61 were used to evaluate the performance of the PF and PAR algorithms. Partial k -space data were extracted from full k -space as required for each of the algorithms. The images were reconstructed according to PF and PAR algorithms and then compared with the image reconstructed using full k -space. At full k -space dimensions, EPRI spatial resolution is $425 \mu\text{m}$ where the smallest gap of $300 \mu\text{m}$ in the phantom is discernible. Examination of this narrow gap served as a measure to assess the resolution loss by PF and PAR algorithms evaluated.

3.2.1 PF algorithms using Hermitian symmetry—Four PF algorithms were evaluated using a rectangular region of k -space covering about 64% of the full k -space: (a) replacing unknown data by complex conjugates; (b) phase correction applied to the conjugated data in (a); (c) homodyne reconstruction; and (d) POCS algorithm to estimate the unknown data. The symmetric k -space region (rectangular strip covering 28%) at low spatial frequencies (Fig. 1B) was used for phase estimation in image space by inverse DFT (Fig. 1S). These phases were used to estimate the k -space values that were not collected according to the procedures described for each of the above PF algorithms respectively. The POCS algorithm described in PF algorithms section was used without a merging filter. The images reconstructed after inverse DFT are shown in Figs. 1 (1I–1L). The absolute differences of these images to the image reconstructed using full k -space (Fig. 1G) are also shown in Fig. 1 (1N–1Q). A smooth phase roll diagonally through the image between π and $-\pi$ is observed in the image (Fig. 1R). Except for simple conjugation (a), all the PF algorithms (b–d) have performed satisfactorily at approximately equal RMS errors (Fig. 1).

3.2.2 PAR algorithm using elliptical k -space region—The choice of partial k -space region used for reconstruction plays critical role in the quality of image since the low frequency region of k -space contributes to image intensities while the high frequencies contribute to spatial resolution. Any discontinuities in low spatial frequency regions lead to intense ghosts and blurring in the image whereas this effect is negligible for the discontinuities in high frequency regions. The success of PF algorithms evaluated above may partly be because of contiguous k -space region including most of the low spatial frequencies. Therefore, it is critical to optimize the geometry of the k -space to be used for partial acquisition strategies. The partial k -space covering a circular region enclosed within the square of a Cartesian grid saves about 21% ($\approx 1 - \pi/4$) of acquisition time. In cases where the k -space energy is distributed in rectangular region, an enclosing ellipse may be considered instead of a circle to further optimize the acquisition. Radii of the ellipse can be calculated from the distribution k -space signal magnitudes. The k -space points having absolute value above a minimum threshold level were used to determine the aspect ratio of ellipse where the threshold was set at mean absolute value of all k -space points. The radii of ellipse were considered as proportional to standard deviations of pixel coordinates above the threshold along each of the respective axes. The radii of the ellipse at two and half standard deviations (2.5σ) covered 42% of the k -space (Fig. 2A). The image reconstructed by zero filling outside the elliptical region (Fig. 2E) and its difference (Fig. 2I) to the image reconstructed from full k -space (Fig. 2H) indicate that this algorithm outperforms the PF

algorithms even at 42% of k -space usage. Increasing the radii of ellipse to 3σ , (including about 60% of the k -space) mainly improved the resolution of the image (Fig. 4). A graph of the percentage of k -space energy covered versus the percent of k -space acquired (Fig. 3) indicates that 60% of k -space coverage includes about 86% of the k -space energy. These observations point out that: (a) the overall k -space energy contributes to the reconstructed image signal level; (b) high spatial frequencies improving the resolution can be traded for scan speed; (c) The aspect ratio of ellipse may be approximated to 1 or can be determined by a coarse scouting scan.

3.2.3 The extent of resolution loss in elliptical acquisition—Truncation of the high spatial frequency region of k -space is expected to decrease the resolution since it is inversely proportional to k -space matrix size in SPI mode as given by the relationship (2),

$$\text{Resolution} = 2\pi / (\gamma_e \Delta G N \tau) \quad [1]$$

where γ_e the electron gyromagnetic ratio, G is the gradient increment step, N is number of gradient steps and τ is the delay from the pulse. Even when there is no reduction in N , the circular k -space selection eliminates the corner regions between the circle and the square enclosing the circle. The differences in the resolutions of the reconstructed images from these two geometries were discussed by Bernstein *et al.* (32). These authors demonstrated resolution improvement by including the corners but point out that it requires an interpolation method such as zero-filled reconstruction. Besides, this improvement has angular dependence and they point out that circular geometry at a higher diameter is preferable for uniform resolution at improved SNR.

For comparison of resolutions, full and PAR reconstructions of the phantom were done without using interpolation. The intensity profiles of a line across the comb-like region of the image were examined to identify the overlaps between adjacent gaps as a function of percentage of k -space (Fig 4). The overlaps are clearly visible at 10%. The RMS difference between PAR and full k -space images decreases at much smaller rate above 20% of k -space (Fig. 4F). The loss of resolution at 10% and 20% of k -space result from the reduction of k -space matrix sizes where both the diameters of the ellipse were less than N .

3.3 PAR for 3D EPRI in vivo

The extension of PAR algorithm described above to three dimensions doubles the scan speed significantly. An enclosing sphere of a diameter equal to the side of the cube includes about 52% ($\pi/6$) of the volume and an ellipsoid having major axis equal to the side of the cube covers even smaller volume. The aspect ratios of ellipsoid are estimated from k -spaces measured at lower dimensions instead of a 3D space to save time. For example, if x , y , and z are principal axes of the ellipsoid, 2D k -spaces are acquired in xy and xz planes and standard deviations of the k -space intensity distribution along x , y and z axes are calculated to determine the aspect ratios of y/x and z/x . Alternatively, three mutually perpendicular 1D k -space acquisitions would serve the same purpose.

An EPRI k -space acquired for 3D imaging of a mouse tumor *in vivo* at a matrix size of $19 \times 19 \times 19$ was used to examine the performance of this PAR algorithm. The aspect ratios of ellipsoid were determined as described above. The percentage of k -space volume enclosed by ellipsoid was calculated in multiples of standard deviation along x -axis, maintaining the aspect ratios of ellipsoid. The graph of k -space energy vs. volume enclosed within the ellipsoid at different diameters was found to be similar to the one obtained for 2D k -space (Fig. 5).

The images were reconstructed by zero-filling the k -space outside the enclosing ellipsoid followed by the calculation of the magnitude of inverse DFT. The comparison of the images reconstructed using different ellipsoid diameters covering 10%, 35%, 52%, 69% and 100% of the k -space data was done using the middle slice (xy plane) for convenience as shown in Fig. 6. Comparison of the image from 100% of k -space with the ones from PAR algorithm indicates that the broad shape of the mouse leg is apparent even by using 10% of the k -space, but finer details appear at 35%. Blurring is greatly reduced when 52% and 69% of k -space is used because of inclusion of higher spatial frequency regions.

3.4 An *in vivo* Oximetry example using partial k -space

For oximetric calculations, k -space data are required at multiple encoding time points on the FID for the assessment of T_2^* . In SPI mode, the FOV is inversely proportional to the delay time τ . The k -space signal magnitudes and FOV are higher at short τ values than those at long τ values, and k -space dimensions increase with delay time. The partial k -space covered at shortest τ value becomes a limiting factor of k -space dimensions in oximetric studies since the matrix size is unchanged throughout the scan. In other words, k -space truncation is done at the earliest FID point used for T_2^* estimation rather than the longer τ values.

We report here the practical utility of PAR in routine 3D *in vivo* animal imaging for oximetry. For the regular T_2^* weighted oximetry using the SPI modality (19), we perform three interleaved measurements with three different gradient steps and consider 12 images (4 each from the three experiments) at different gradients as a function of delay from the pulse, such that the images have almost similar intrinsic resolution for performing *in vivo* oximetry. Note that in SPI, the images show a progressive zoom-in (progressively decreasing FOV) which is a function of the gradient steps and the delay being given by

$$\text{FOV(SPI)} = 2\pi / (\gamma_e \Delta G \tau) \quad [2]$$

By keeping the product $G\tau$ nearly constant one can retain near uniform resolution in a three-gradient interleaved SPI experiment. In performing oximetry, the 12 images as a function of delay are made to occupy identical volume using chirp z transform (16) so that the voxels in the images correspond to identical locations in the subject and are only subject to T_2^* decay as a function of delay from the pulse. To simulate the PAR mode, the full k -space data from a three-gradient interleaved SPI experiment was truncated using a spherical cut-out of all the 12 data sets followed by chirp z transform. Once the 12 images were generated from the partial k data, T_2^* and pO_2 values were estimated following the least squares fit and pO_2 calculation using procedures described previously (1). Fig. 7A shows the

surface-rendered 3D image of a tumor-bearing leg of a C3H mouse processed from the full and partial k data sets with the partial k datasets corresponding to a spherical cut-out retaining 54% of the total k -space data. The right image from the partial k -data does show a reduction in the resolution by way of smearing the sharp edges seen on the left image from the full k data. Figs. 7B and 7C show the profile of spin concentration and oxygen partial pressure (pO_2) along arbitrarily chosen directions in the 3D spin and oxygen images derived from the partial and full k data sets, and indeed the differences one sees are within the statistical error one obtains from the full k image. In fact spin distribution slices and oxygen map slices obtained from the full and partial k data sets are almost identical and their differences are negligible as shown (along an arbitrary direction) in Figs. 7D and 7E. It is therefore encouraging that nearly 50% of the measurement time can be saved in exchange for higher temporal resolution and dynamics of the spin and oxygen distribution without compromising the quality of the results using the PAR strategy described here.

3.5 Partial k -space acquisition of a phantom

Although partial elliptical k -space increases scan speed, acute deviation from circular geometry may lead to unequal blurring and variable distortions along different directions of the image. An actual partial k -space acquisition was made as opposed to extraction of partial data from full k -space described in previous sections in an effort to confirm the validity of simulations since the k -space scanning trajectory alters during elliptical acquisition. Full and partial k -space data were acquired by 2D imaging of a five tube phantom at maximum gradient ($G_{\max}$) = 10 mT/m (Fig. 8). In addition, partial k -space acquisitions were done in circular and elliptical settings at three $G_{\max}$ values of 12, 10, and 8 mT/m. The k -space regions excluded at higher $G_{\max}$ values by partial acquisition have lower energies than those excluded in the cases of lower $G_{\max}$. Comparison of circular k -space at 12 mT/m (Fig. 8B) with 8 mT/m (Fig. 8D) reveals larger magnitude changes at the edges of circle for 8mT/m than at 12 mT/m. This results in higher ringing artifact in the reconstructed image (Fig. 8H) at 8 mT/m. Truncation of k -space leads to Gibbs ringing when the intensities at the periphery are substantial. This can be reduced to a large extent by the use of radial windowing and zero-filling as has been discussed by Bernstein et al.(32). However, in SPI mode it is possible to select a longer delay time to avoid truncation of high energy region of the k -space. Further the FOV of the image varies with $G_{\max}$ as shown in Figs. 8F – 8H.

Partial k -space enclosed within an ellipse having aspect ratio of 0.5 is shown in Figs. 8I-8K. Exclusion of k -space at the top and bottom as shown in Fig. 8K lead to vertical distortion in the reconstructed image as shown in Fig. 8N. However, these distortions are not apparent at 12 mT/m (Fig. 8L) because of the low energies of excluded k -space regions at the top and bottom. Therefore, either the aspect ratios should be close to 1 or k -space signals excluded need to be negligible for this PAR algorithm.

3.6 Scan time constrained ellipsoid dimensions

We considered scan speed as the determinant for the selection of partial k -space that best suits to the application. In this scheme, the percentage of total k -space to be acquired is given by an input parameter to constrain the radii of the ellipsoid. Given that the fraction of

the total k -space to be scanned is f , and ellipsoid aspect ratios are $k_y = r_y/r_x$ and $k_z = r_z/r_x$, the radius r_x of the ellipsoid in points is given by

$$r_x = 0.5 \left[6 f N^3 / (\pi k_y k_z) \right]^{1/3} \quad [3]$$

where N is the number of k -space points of a side of the cube. The radii r_y and r_z are calculated from k_y , k_z and r_x values. For 2D acquisition, the value of r_x in points is given by

$$r_x = \left[f N^2 / (\pi k_y) \right]^{1/2} \quad [4]$$

Actual percent of k -space acquired may be slightly different depending on the number of k -space points included inside the ellipsoid or ellipse defined by these radii.

4. Conclusions

In imaging live objects with infused spin probes which undergo concentration change due to fast pharmacokinetics and or renal elimination, reduced k -space strategies will be useful for investigating fast dynamics with improved temporal resolution. A common problem in PF algorithms is the estimation of phase correction. Inaccurate estimates lead to artifacts arising from abrupt changes in k -space and blurring. The PAR algorithm enclosing high-energy part of the 3D k -space by an ellipsoid or sphere is found to be robust for EPRI. This procedure almost doubles the scan speed. Combining PF algorithms utilizing the Hermitian symmetry of the k -space with this procedure can further improve the measurement time by an additional factor of 2, enabling better temporal resolution or improved SNR in a given measurement time.

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Figure 1.

Performance of PF algorithms in EPRI. A–F: *k*-space signal magnitudes of a 2D image. A. Full *k*-space matrix (61 × 61), B. Symmetric strip used for phase estimates (17 × 61), C–F. Left 39 lines (64%) are known data from A and the remaining *k*-space data were estimated by: (C) Hermitian symmetry, (D) phase correction applied to estimates in C, and (F) POCS. Weighted *k*-space for homodyne reconstruction is shown in E. G–L: The images reconstructed from A–F respectively. The magnitude of differences of images H – L from G are shown in M – Q (RMS differences = 2.64, 2.57, 1.09, 1.46 and 1.58 × 10⁻⁶ respectively). R – W: Phases of images G – L (grays black to white are phase angles -π to π respectively).

Figure 2.

Performance of PAR algorithm using elliptical k -space region. A–D: Absolute values of 2D k -space in elliptical regions covering 42%, 60%, 86%, and full k -space. E–H: Images reconstructed from A–D respectively by zero-filling the uncollected data. I–K: differences of E–G respectively from H. The matrix size of k -space is 61×61 . Gradient strength = 40 mT/m, and FOV=2.59 cm. L: RMS intensity difference per pixel from image H as a function of percent k -space used. O – P: Real part of the images E – H indicating the presence of a slow phase variation.

Figure 3. Percentage of k -space energy covered versus percentage area of k -space used for 2D face-phantom EPRI data.

Figure 4.

Intensity profiles of row 39 across comb-like region in 2D phantom image using full and elliptical k -space. The matrix size of k -space is 61×61 . Gradient strength = 40 mT/m, FOV=2.59 cm. A–E: Elliptical regions covering 10%, 20%, 42%, 60% and 86%. Dashed and solid lines indicate partial and full k -space profiles respectively. F: RMS intensity difference per pixel of row 39 as a function of percent k -space used. G: Row 39 is indicated by dashed line in phantom image.

Figure 5.
Percentage of k -space energy covered versus percentage volume of k -space used for 3D EPRI data *in vivo*.

Figure 6.

Performance of PAR algorithm using ellipsoid region of 3D EPRI k -space data (matrix size = $19 \times 19 \times 19$). Images were reconstructed using 10%, 35%, 52%, and 69% enclosed by ellipsoids and full k -space. A–E: Absolute values of middle-slice (slice 10) of k -space. F – J: Slice 10 of the reconstructed image. Bright oval shape is mouse leg bearing tumor. The intensity variations in spin density indicate the distribution of spin probe infused. K – N: differences of the images F – I respectively from the full k image J. Reconstruction was done by zero-filling and inverse FFT. Gradient strength = 14 mT/m. O: The positioning of the mouse tumor leg in the resonator.

Figure 7.

A. Surface rendered 3D spin image of a tumor-bearing leg of C3H mouse with full k data (left) and 54% k-data (right). B. spin images from a sagittal slice passing through the middle of the tumor: full k (left) and partial k (right). C. pO₂ map from the slices of B for full k (left) and partial k (right). D. spin density along the dotted line in (B): full k (blue) and partial k (red) and E. pO₂ along the dotted line in (C) full k (blue) and partial k (red).

Figure 8.

Full and Partial 2D k -space measurements on a five tube phantom. A: Magnitude values of full k -space points (21×21). B–D: Partial k -space (79% of total space) acquired within the enclosed circle shown by white line at 12, 10, and 8 mT/m respectively. E – H: images reconstructed from respective k -spaces shown in A–D at matrix size 256×256 . I–K: partial k -space acquired within elliptical enclosure (59% of total space shown by white line) where the aspect ratio of minor to major axes of the ellipse = 0.5 at 12, 10, and 8 mT/m respectively. L – N: images reconstructed from the respective k -spaces shown in I–K (256×256). O: Schematic description of phantom viewed from top. At lowest maximum gradient magnitude vertical distortion of tubes is apparent (see N) for aspect = 0.5 since high energy k -space is truncated asymmetrically along major and minor axes.