

Rapid Magnetic Resonance Imaging Using Undersampled Projection-Onto-Convex-Sets Reconstruction

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ABSTRACT

Scan time reduction is important in clinical magnetic resonance imaging (MRI). Partial Fourier data acquisitions rely on the conjugate symmetry of Hermitian data, allowing for shorter scan times due to fewer phase-encoding steps needed during the MRI data acquisition. Estimation of the missing \mathbf{k} -space MR data is then usually accomplished by direct conjugate synthesis (e.g., homodyne reconstruction) or by iterative constraint-based algorithms (e.g., projection-onto-convex sets, POCS). Faster image acquisition has been achieved by combining partial Fourier acquisitions with parallel MRI. In this study, an iterative partial Fourier reconstruction method based on POCS technique is presented for sharply decreasing MRI scan times. It is achieved by asymmetrically undersampling \mathbf{k} -space MR data acquired using a multi-element MRI receive coil. The results obtained from experimental models (phantoms) and humans shows that the proposed method can rapidly produce high-quality MR images with high acceleration rates.

Keywords: MRI, Image Reconstruction, Rapid MRI Data Acquisition, POCS, Parallel Imaging

I. INTRODUCTION

Magnetic resonance imaging (MRI) is now a routine medical imaging modality for clinical diagnosis. It is considered a minimally invasive approach in that harmful ionizing x-ray radiation is not used and its adverse reactions are negligible. Technically, however, in MRI, the imaging time, the signal-to-noise ratio (SNR), the spatial and temporal resolutions are inherently interdependent. The fixed relationship between these properties causes trade-offs that improving one characteristic would worsen the others. And, despite its tremendous clinical efficacy and exquisite sensitivity to subtle changes in living tissues, MRI is still considered a relatively slow modality for many clinical applications. To address this important limitation, parallel imaging (PI)

techniques have been proposed in MRI [1-6] with recent remarkable results in clinical applications [7-15] such as cardiac, abdomen, thorax and brain imaging. In MRI, the emitted echoes from the imaged object are detected and collected in the spatial frequency domain, *i.e.*, Fourier domain known as \mathbf{k} -space, by means of a radio frequency (RF) receiver coil surrounding the object. PI reduces the total scan time and/or increases the spatiotemporal resolution by using multi-element phased-array (PA) receiver RF coils. In particular, PI exploits the difference in reception sensitivities (due to different in location, gain, direction, size, etc.) between individual coil elements (N_c) in a receive array to reduce the high number (usually > 64) of the phase-encoding steps required in MR imaging. The detected signal by the i^{th} element now can be described as:

$$G_i(\mathbf{k}) = \sum_{\mathbf{r}} s_i(\mathbf{r}) e^{-j\mathbf{k}\mathbf{r}} f(\mathbf{r}) + n_i(\mathbf{k})$$

$$i = 1, 2, \dots, N_c$$

in which \mathbf{r} , the position vector, $s_i(\mathbf{r})$, the coil sensitivities (coil magnetic field maps), \mathbf{k} , \mathbf{k} -space samples vector, $n_i(\mathbf{k})$, the coil noise, and $f(\mathbf{r})$ is the MR image of the object to be estimated. Therefore, the image reconstruction process is an inverse problem usually solved with least squares methods.

PI was originally conceived [1-3] as a means of ultra-fast imaging using only a single MR phase-encoded readout echo, replacing the traditional gradient fields' phase-encoding steps entirely with spatial-encoding using the coil sensitivities of a large number of small elements. However, it soon became apparent that there were fundamental as well as practical limitations to the effective number of spatial-encodes that were possible. This led to more realistic implementations [4-15] that used a reduced number of phase-encoding steps. Currently, the newest generation of MRI scanners typically provide 32 independent receiver channels, which theoretically allow a $32\times$ increased image acquisition speed compared to traditional MR systems without PI environment.

Over the past years, great progress in the development of PI methods has taken place, thereby producing a multitude of different and somewhat related parallel imaging reconstruction techniques and strategies. Currently, the most well-known are SMASH,[7] SENSE,[12] and GRAPPA.[16] However, various other techniques, such as AUTO-SMASH, [17] VD-AUTO-SMASH,[18] GENERALIZED SMASH,[11] MSENSE,[19] PILS,[20] and SPACE RIP [21] have also been proposed. All these techniques require additional coil sensitivity information to eliminate the effect of undersampling the \mathbf{k} -space. This sensitivity information can be derived either once during the patient setup by means of a prescan or by means of a few additionally acquired \mathbf{k} -space lines for every

subsequent PI experiment (autocalibration), or some combination of the two. The present PI reconstruction methods can roughly be classified into two groups. Those in which the reconstruction takes place in image space (*e.g.*, SENSE, PILS) consist of an unfolding or inverse procedure and those in which the reconstruction procedure is done in \mathbf{k} -space (*eg.*, SMASH, GRAPPA), consist of a calculation of missing \mathbf{k} -space data. However, hybrid techniques like SPACE RIP are also conceivable.

The partial Fourier method [22-24] is another approach for decreasing total scan time, which reduces the amount of acquired data by asymmetrically truncating the peripheral portions of \mathbf{k} -space in either the phase-encoding direction (to reduce Nphase) [23] or the frequency encoding direction (to reduce TR) [24], or both. Zero-filling (ZF) [25], homodyne [26] or projection onto convex sets (POCS) [27,28,30-31] techniques may be used to reconstruct these images.

Even faster image acquisition can be achieved by combining partial \mathbf{k} -space acquisitions with parallel imaging. For example, the POCS formalism has been used to reconstruct sensitivity-encoded MR data in an iterative POCSSENSE procedure. [29] However, we hypothesize that the \mathbf{k} -space-based GRAPPA or SMASH parallel imaging formalisms are more compatible with partial Fourier methods because the synthesis of the full-FOV image can be done in \mathbf{k} -space prior to homodyne or POCS reconstruction. Unlike with SENSE and SMASH, detailed accurate sensitivity maps are not needed prior to GRAPPA reconstruction. [16] This coil sensitivity information is obtained from additional central \mathbf{k} -space lines (*i.e.*, auto-calibration signals, ACS) acquired during the actual scan and are incorporated in the image reconstruction for higher SNR. Here we propose combined GRAPPA and POCS techniques that allow simple and computationally efficient inclusion of phase and data-consistency constraints in image

reconstruction in order to both improve image quality and to achieve higher acceleration factors. As with GRAPPA, these methods provide unaliased images from each component coil prior to image combination resulting in high SNR and better image quality. We evaluate the performance of this method using experimental models and *in vivo* data.

II. METHODS AND MATERIAL

Fully sampled k -space data sets were collected on a 3.0 T clinical MR scanner (MR 750; GE Healthcare, Waukesha, WI) using a vendor-supplied quality assurance phantom and volunteers. Written consents were obtained from volunteers according to our institutional review board. Images were acquired using a 4-element torso array coil. A 2D fast gradient-recalled echo sequence (TR/TE/flip = 8.1ms /3.2ms /60°; 22-cm FOV; 256 × 256 acquisition) was used to image the phantom. A 2D fast spin echo sequence (TR/TE/flip = 217ms /20ms /45°; 32-cm × 16-cm FOV; 256 × 256 acquisition) was used to image the legs. Partial k -space parallel acquisitions with accelerations rates of $R = 2, 3, 4,$ and 5 were emulated by acquiring only every second or third phase-encoding line of one side of k -space (i.e., positive k_y) and removing all but the central 32, 16, or 8 phase-encoding lines of the k -space. These central lines were used as auto-calibration signals (ACS) for GRAPPA and for estimating the low-resolution phase map in POCS

reconstruction (Fig. 1). Three methods of image reconstruction on the partially sampled k -space data were investigated: (i) conventional GRAPPA reconstruction (according to Ref. [16]) (ii) POCS+GRAPPA: POCS k -space reconstruction (according to Ref. [28]) was first performed followed by GRAPPA, and (iii) GRAPOCS (Fig. 2): GRAPPA k -space reconstruction followed by POCS [30,31]. A conventional sum-of-squares (SOS) reconstruction method was also performed on fully sampled data from all coil elements and used as reference.

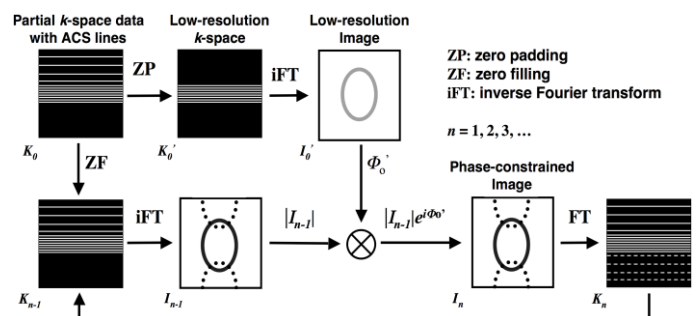


Figure 1: POCS algorithm on partially undersampled k -space data. K_0 : Original undersampled k -space data, K_0' : central zone of k -space, I_0 : initial reconstructed image from which the low-frequency phase map is extracted (Φ_0'). K_{n-1} : iterative estimations of the full k -space, used to generate an estimate of the reconstructed image (I_{n-1}). I_n : iterative phase-corrected image, K_n : new estimated k -space obtained by substituting missing data in the original undersampled k -space data. The process is repeated until a convergence criterion is met.

III. RESULTS AND DISCUSSION

Figs. 3 and 4 show the phantom and axial images in

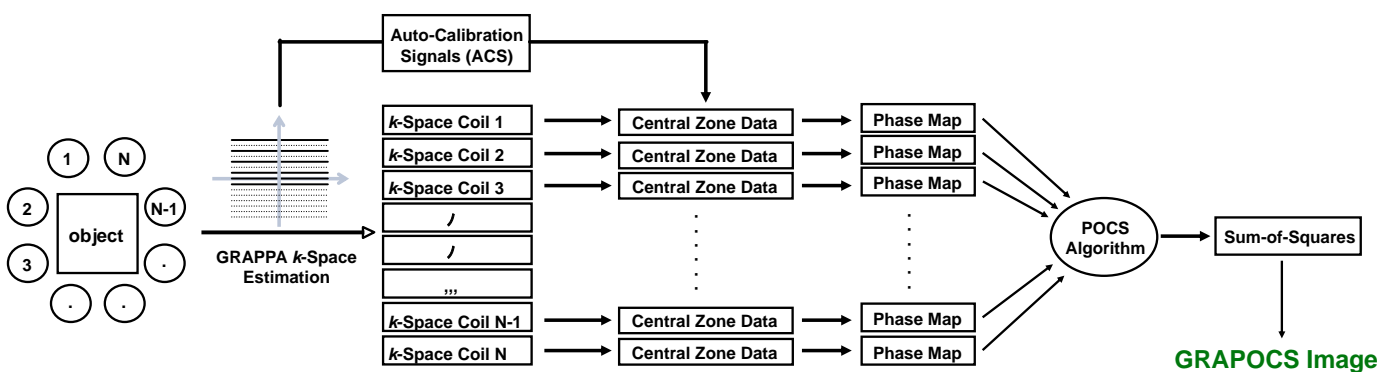


Figure 2: The GRAPOCS reconstruction algorithm for partially parallel imaging. Accelerated partial Fourier MR data are acquired in parallel using a N -element coil. Conventional GRAPPA [16] is first applied to obtain the full k -spaces from all coil elements. POCS algorithm (Figure 1) [28] with phase- and data-consistency constraints is then applied followed by a sum-of-squares combination.

the legs using the four reconstruction methods described above. In each case, the SOS images had the best quality and highest signal-to-noise ratios, as expected, but had no scan time reduction effect ($R = 1$). For the same scan time reduction factor, R , the GRAPPA images appeared blurred compared to other methods. The POCS-based reconstructed images had higher resolution and low residual aliasing, particularly the preferable GRAPOCS implementation order (compare Fig. 1c and 1d). For higher acceleration factors, the improvement in image quality was even greater for GRAPOCS, particularly when $R >$ number of coil elements (Fig. 4, last row).

parallel imaging. Because POCS reconstructs complex data and includes phase information, it is beneficial to use this method in conjunction with auto-calibrated parallel imaging methods. Here, we demonstrated the use of POCS with GRAPPA. As with GRAPPA, the proposed GRAPOCS method provides unaliased images from each component coil prior to image combination but results in better image quality for greater accelerations rates.

The proposed GRAPOCS method is a data processing technique that can directly work on any MR scanner equipped with PI capability. Like many other PI methods, it is a core development applicable for most MR sequences and with most clinical applications regardless of field strength or vendor design of the scanner hardware. With this method, neither the desirable image contrast will be altered nor shorter repetition times (i.e., higher gradient switching rates) will be necessary. Our preliminary results from different objects/regions (a quality assurance phantom and legs of a volunteer) has shown promise in imaging various human organs. These methods can be used in other MRI applications (e.g., knee, brain, heart, torso, abdomen, etc.). Because the power spectrum (k -space power distribution) varies between different regions/objects, larger number of k -space central lines can be acquired to produce higher resolution phase maps for the POCS algorithm, therefore further enhancing the GRAPOCS technique for various MR clinical applications.

Like any other PI methods [7-21], our GRAPOCS methods shift time from acquisition to reconstruction. This distinction is not terribly relevant since in MRI it is most critical to minimize scan time for the patient. For example, one would like to minimize the time of breath hold when cardiac imaging [32]. However, from the diagnosis point of view, the reconstruction time can be important too. Typically, vendors provide a workstation configuration - i.e., quad-core CPU

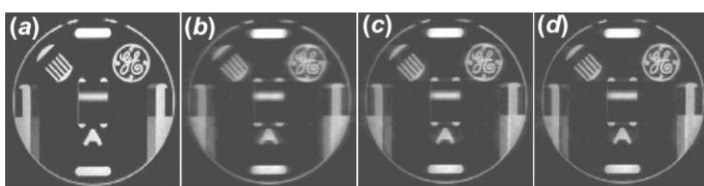


Figure 3: Phantom results from (a) SoS ($R = 1$), and (b) GRAPPA, (c) POCS+GRAPPA, and (d) GRAPOCS reconstructions with acceleration rate of $R = 5$.

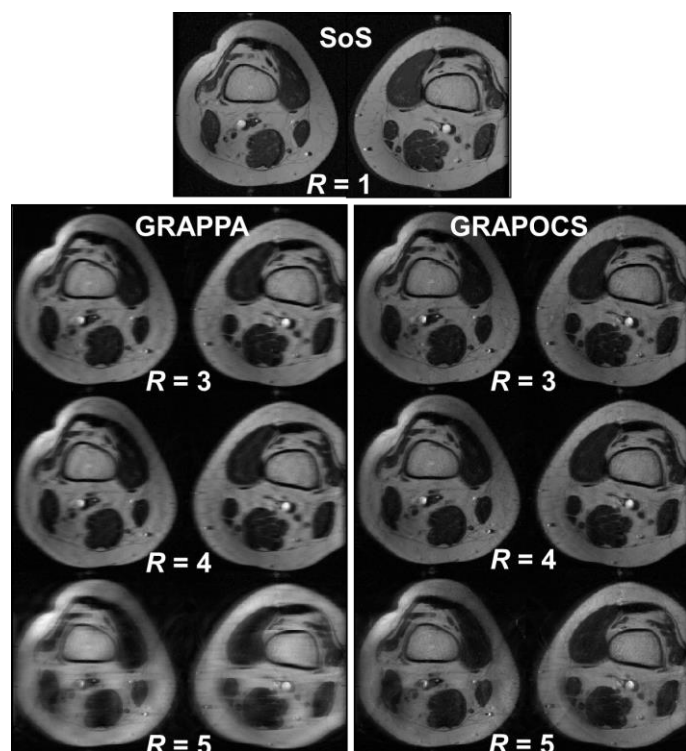


Figure 4: Axial images from a volunteer's legs using SoS, GRAPPA, and GRAPOCS reconstructions with simulated $R = 3, 4, 5$. Data acquisition used a 4-element receive coil. GRAPOCS produces better images.

To accelerate MR scans, it is often desirable to use partial Fourier acquisition in conjunction with

processor of 2-3 GHz - for reconstruction. Reconstruction time can then be minimized by using faster reconstruction computers to produce GRAPOCS images in a few seconds.

IV. CONCLUSION

We have combined two reconstruction methods each of which is known to be effective for scan time reduction. The inherent ACS lines acquired in GRAPPA are synergistic with the need for estimation of a phase map in POCS. These proposed methods can be used in various MRI clinical applications rapidly acquiring data and producing high quality images.

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VI. REFERENCES

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