A k-space implementation for image support minimization to improve parallel imaging performance in dynamic imaging

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Introduction

A fundamental step in the *k-t* SENSE [1] method is minimizing the support of the image in *x-f* space and hence improving reconstruction performance. This idea can be adopted in *k-t* space and used with a *k*-space based reconstruction method. For a dynamic *k*-space sequence, it is easy to calculate the *k*-space mean along the time direction. A new sequence can be generated by subtracting this mean from the original *k*-space time frames. It is often true for dynamic images that the new *k*-space sequence correponds to images that have reduced support (in image space). We postulate that this *k*-space may be better for parallel imaging techniques. As a specific example, we show that GRAPPA [2] can generate better results with the residual *k*-space sequence. Experiments on caridac MRI lend support to this statement.

Method

As a specific example, the *k-t* space acquisition scheme described in [3] is applied here. For this time interleaved *k*-space we average by simply adding acquired points along the time direction and then dividing by the number of these points. Then the residual *k*-space is calculated by subtraction of this mean *k*-space from each *k*-space frame. GRAPPA is then applied to the residual *k*-space. Finally, this result and the mean *k*-space are added back together to obtain the full reconstructed *k-t* space. To further take advantage of the reduced support in image space, one step of the POCS [4] idea can be adopted here. Based on the initial reconstruction using GRAPPA, the static region with very low intensity can be easily found by using a threshold of one standard deviation along the time direction. To avoid the influence of intensity in the standard deviation map, it can be normalized by dividing by the mean image generated from the mean *k*-space. One should notice this static region normally does not include all the background region of the original images, because the background region in the original images contains only noise and has a large standard deviation along the time direction. Figure 1D) and 2 D) show the static regions (white) generated in our experiments. This region can be set to zero to reduce the noise and generate a set of modified images. Those modified images can be projected back into *k*-space to pick up the acquired data and generate a modified full *k*-space. Sum of squares is then applied to the full *k*-space to reconstruct the final image sequence.

Results

Oblique cardiac images were collected by a 1.5T GE system (FOV 280 mm, matrix 160×120, TR 4510 ms, TE 2204 ms, flip angle 45°, Slice thickness 6 mm, number of averages 2) through fast imaging employing steady-state acquisition (FIESTA) with a GE 4-channel cardiac coil. Figure 3 shows the results for axial cardiac images collected by the same 1.5T GE system (FOV 240 mm, matrix 192×256, TR 4530 ms, TE 1704 ms, flip angle 45°, Slice thickness 5 mm, number of averages 1). In both cases, breath-holds ranged from 10 - 20 seconds. There are 20 images per heartbeat. Table 1 allows for the comparison of relative errors from GRAPPA using the original *k*-space, the modified *k*-space, and POCS using the modified *k*-space (the reduction factor is 4). To find the relative error, pseudo-partial *k*-space was used. This means that the reduction factor is 1 for acquisition, but only a quarter of the phase encoding lines, with 24 fully sampled lines in the center band of *k*-space, were used for reconstruction. The relative error is defined as the relative difference of the energy between the reference image and the reconstructed image, divided by the energy of the reference image. The numbers given in Table 1 are the mean relative errors over each time frame. The dynamic region is defined as the region around the heart. Figure 1 and 2 show various results using the first frame of an oblique image sequence and an axial image sequence. A) is the reference image generated from the full *k*-space, B) is the image reconstructed using GRAPPA with the original *k*-space, C) is the image reconstructed using GRAPPA with the residual *k*-space and then adding the mean *k*-space, D) is the static region (white) generated from the initial reconstruction, E) is the image reconstructed using GRAPPA with the residual *k*-space and POCS.

		Original *k*	Residual *k*	Residual *k* & POCS
Oblique	Dynamic region	12.72%	2.68%	2.55%
	Whole image	15.13%	5.26%	5.04%
Axial	Dynamic region	24.50%	13.05%	3.90%
	Whole image	21.21%	8.49%	8.04%

Table 1. Relative errors using different methods.

References

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[3] Huang F, *et al*, the 2nd parallel imaging workshop, Zurich, p53, 2004.
[4] Kholmovski EG, *et al*., ISMRM 2002, p194.

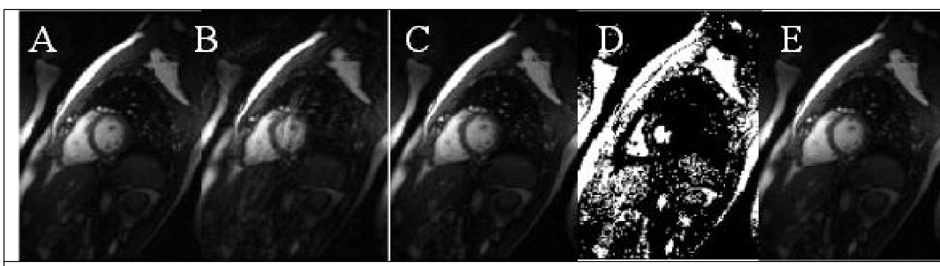

Figure 1. Various results using the first frame of the oblique image sequence.

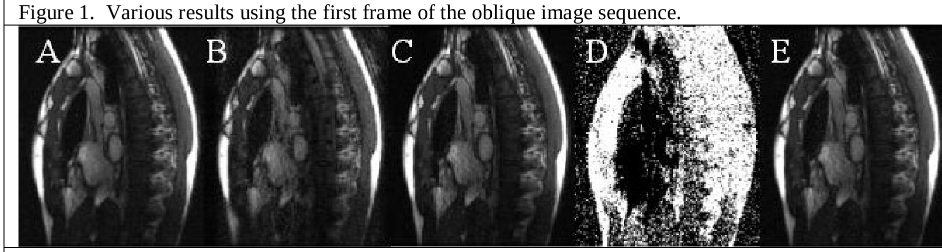

Figure 2. Various results using the first frame of the axial image sequence.

Discussion

From these experiments it can be seen that the residual *k*-space may be used by GRAPPA to improve reconstruction quality. The reason is that the energy in the residual *k*-space is not dominated by the central lines (which tend to cancel more after subtraction), hence the calculated weights are more suitable for the high frequency *k*-space region. The POCS idea can be applied to further take advantage of reduced support and hence reduce noise. This method only modifies the *k*-space before and after the original reconstruction method (GRAPPA in this example), with no need for sensitivity maps, hence it is easy to implement. There are no time consuming calculations for *k*-space modification (only addition, subtraction and division by a constant), and there are only two more fast Fourier transforms needed for POCS, hence the method is fast. Moreover, artifacts and noise can be dramatically reduced by this simple, cheap process. The residual *k*-space concept may also be applied to other parallel imaging techniques. We have also done experiments in which *k-t* GRAPPA was improved.