MRI with Accelerated Multi-Coil Compressed Sensing

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Introduction

Parallel imaging methods such as SENSE [1] reduce acquisition time at the expense of aliasing artifacts and image SNR. Compressed sensing (CS), surprisingly, states that we may sample a compressible signal at a rate lower than Nyquist's [2,3,4] and achieve exact reconstruction. If an image may be compressed retaining only K coefficients in some basis, we call the image K-sparse. The K coefficients, according to CS theory, may be recovered with M ~ cK measurements in a basis that is incoherent with the compression basis. Generally, the constant c is O(1) and depends weakly on the size of the image. For example, a measurement matrix of 15,000 k-space points chosen at random reconstructs a 256x256 Shepp-Logan phantom (K=3749 in a basis of Haar wavelets) for an acceleration factor R=4. Distributed compressed sensing (DCS) [5] extends CS to partially-correlated multiple receivers with another surprising result: in the limit of infinitely many receivers, the number of measurements necessary for exact reconstruction shrinks to M = K+1 if all J signals have identical support (with different coefficients). In the previous example, for an infinite number of receivers, only M = K+1 = 3750 k-space points are required (R=17). Simulations [5] reveal that "infinitely" many receivers is roughly J~32 (for Gaussian noise sources), within the reach of current MR scanners. We develop the application of distributed compressed sensing to MR parallel imaging.

Methods

Each coil of a multi-channel MR system acquires data encoded by phase, frequency, and coil spatial sensitivity. This relation is given by: $y_j = \Phi_{FT}\Phi_{CSP,j}x$ where y_j is a vector of k-space measurements from the j-th coil, x is the object, Φ_{FT} the Fourier transform and $\Phi_{CSP,j}$ coil sensitivity weighting of the j-th coil. If *x* is compressible, we can introduce a basis Ψ, incoherent with Φ_{FT} and $\Phi_{CSP,j}$: $y_j = \Phi_{FT}\Phi_{CSP,j}\Psi s = \Phi_M s$ The number of measurements Y={y_1,…,y_J} may be greatly reduced from the Nyquist rate, if s has few non-zero elements in the sparse domain Ψ. The task is to recover *s*, the coefficients of *x* projected into the basis Ψ. In general this optimization problem is combinatorial and NP-hard [3,4]. Tropp, et al. [6] introduced Simultaneous Orthogonal Matching Pursuit (SOMP), a greedy solver to determine *s*. We adapt SOMP to parallel imaging with J coils; briefly, SOMP determines iteratively the best coefficient of *s* to estimate, allowing each coil to "vote" equally. For MR, we changed the algorithm such that each coil's vote for the best coefficient is weighted by the sensitivity function. Once a sufficient number of coefficients are determined, or the difference between measurements and estimate is small, SOMP terminates, producing J estimates of the object X = {x_1,…,x_J}. We combine the estimates by optimal phased array combining [8] to produce the final reconstructed image.

Figure 1: Shepp-Logan phantom, a) original, b) simulated ideal coil image, c) 4000 wavelet coefficient reconstruction by SOMP from 15,000 measurements, d) SOMP reconstruction with 10,000 measurements and e) SOMP reconstruction with 5,000 measurements showing degradation in agreement with theory.

Results

The Shepp-Logan phantom has K=3749 non-zero components under a 3 level Haar wavelet transform. We simulated an 8 channel parallel acquisition by weighting the Shepp-Logan phantom with Biot-Savart coil sensitivities for circular coils arranged regularly on an octagon and optimally overlapped. We assume a 3D acquisition with a 1D inverse FFT along the read direction; arbitrary k-space acquisition patterns may be used. A grid of 25x25 measurements was forced around DC, and the remainder uniformly distributed in a 256^2 matrix. The phantom is shown in Figure 1a), 1b) shows an individual ideal coil image. In DCS, 32 receivers are close to "infinity" for Gaussian noise sources [5]. Even with receiver counts in parallel imaging approaching 128 [7], we begin our exploration with 8 channels to limit computational burden. Figure 1c) is a reconstruction with 15,000 k-space points (R=4.4), and 1d) utilizes 10,000 k-space points (R=6.5). The reconstruction of 1e) uses 5,000 k-space points(R=13), and, as theory predicts, the reconstruction starts to break down as these are too few measurements for the number of receivers and sparsity of the phantom. A phantom image and coil sensitivity profiles were acquired on a 1.5T scanner (GE Healthcare, Waukesha, WI) using an FGRE sequence and an 8 channel cardiac coil. We constrain the reconstruction to 2D after random sampling of the k-space data. The phantom acquisition of Figure 2a) was reconstructed from 15,000 random k-space points; due to computational burden, SOMP was only allowed to recover the first 4,000 wavelet coefficients but the reconstructed image is consistent with the gold standard truncated to 4,000 wavelets. With increased iterations, we expect image quality to approach the gold standard. From the fully sampled k-space data, we simulated an R=4.8 SENSE reconstruction (Figure 2b). Note the aliasing artifact in the center of the phantom. As a gold standard, Figure 2c) shows the fully sampled optimal phased array reconstruction [8].

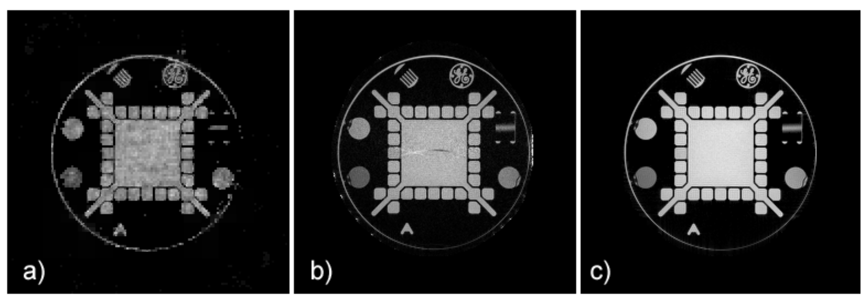

Figure 2: FGRE axial phantom acquisition. a) SOMP reconstruction from 15,000 k-space points of the first 4,000 wavelet coefficients. Reconstruction was restricted to 4,000 wavelets to limit computations, but a faithful reconstruction has been achieved: further iterations will improve convergence. b) 2D SENSE reconstruction with R=4.8 - note aliasing artifact in the center of the phantom. c) Optimal phased array reconstruction with fully sampled k-space.

Discussion

In this work we have exploited the partial correlation of many coils for exact reconstruction of a Shepp-Logan phantom using a novel adaptation of DCS to MR imaging. We have successfully applied this reconstruction algorithm to phantom data demonstrating a speedup factor of 4.4 relative to conventional imaging. Combination of MR-DCS and SENSE is in progress to achieve high speedup factors with limited g-factor impact. Computational burden must be addressed to prove clinical feasibility.

References: 1) Pruessmann et al. MRM 42, 952(1999), 2) Lustig et al. ISMRM 2005, 3) Candes et al., IEEE Trans. Info. Theory 52, 489(2006), 4) Donoho, IEEE Trans. Info. Theory 52, 1289(2006), 5) Duarte et al. Sig., Systems & Computers 2005, 6) Tropp et al., ICASSP 2005, 7) Hardy et al. ISMRM 2007, 8) Roemer et al. MRM 16, 192 (1990).