A Method for Autocalibrating 2-D Accelerated Volumetric Parallel Imaging with Clinically Practical Reconstruction Times

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The scan time for volumetric acquisitions can be significantly shortened by using parallel imaging to reduce the number of acquired phase encodes in both phase-encode directions; previous work has presented volumetric results for 2-D accelerated SENSE (1) and GRAPPA (2). One of the main challenges for 2-D accelerated autocalibrating parallel imaging is the significant increase in computation required to generate and apply volumetric kernel weights for each sampling pattern encountered in k-space. In this work, we present a method for autocalibrating parallel imaging reconstruction that employs a novel calibration procedure to simplify computation requirements for 2-D accelerated volumetric imaging. This calibration procedure combined with a hybrid space application phase (3) forms the basis for the ARC (Autocalibrating Reconstruction for Cartesian sampling) method. Preliminary clinical results using ARC are demonstrated in contrast-enhanced abdominal studies.

Theory The typical autocalibration procedure generates the reconstruction weights w by fitting a 'source' matrix D_s to a 'target' matrix D_t using generalized matrix inversion. By implementing an indirect procedure of using the calibration data to calculate 'correlation values' which are then used to calculate reconstruction weights, ARC reduces the computation involved in generating the reconstruction weights. The generalized matrix inversion approach computes the reconstruction weights as $w = (D_s^H D_s)^{-1} D_s^H D_t$. We rewrite this as $w = C_{ss}^{-1} C_{st}$, where $C_{ss} = D_s^H D_s$ and $C_{st} = D_s^H D_t$. Examination of the matrix multiplications forming C_{ss} and C_{st} reveals that each element in these matrices can be directly computed as a correlation value $c(j_1, k_1; j_2, k_2)$ relating two points in the k-space kernel across coils, as illustrated in Fig 1. For the 3x7x7 kernel and source/target pattern shown in Fig. 2(a) with 8 receiver channels, directly computing $C_{ss} = D_s^H D_s$ and $C_{st} = D_s^H D_t$ by matrix multiplication would require 150,528N complex multiplications, where N is the number of source/target fits in the calibration region and can easily be in the tens of thousands for volumetric calibration. It can be shown that significant overlap in computation exists between correlation values: by capitalizing on this, all necessary correlation values for the source/target pattern in Fig. 2(a) can be computed with less than 10,000N complex multiplications, a 15X improvement. Furthermore, these correlation values can be reused to calculate the weights for other source/target patterns, such as those in Fig. 2(b) and (c). By using correlation values, the data matrices D_s and D_t do not ever need to be formed or multiplied. As these matrices can become very large when 3-D kernels are used, this can also result in significant computation memory and time savings.

Methods Twenty patients were scanned during routine clinical scanning on a 1.5T scanner (Signa® HDx, GE Healthcare, Waukesha, WI) using an 8-channel body array. 3-D T1w contrast-enhanced LAVA (Liver Acquisition with Volume Acceleration) data were acquired with 2-D variable-density acceleration (maximum acceleration of 2 in each phase-encode direction), partial k-space acquisition in k_z and a calibration region of 20x20 phase encodes. Typical imaging parameters were: 320x224, 48 slices (interpolated to 96), slice thickness 4.4 mm, BW ±62 kHz, TE/TR 2.1/4.5 ms, FOV 40 cm, scan time of about 20 seconds depending on patient anatomy.

ARC was implemented online as a host-based reconstruction on a 3.2 GHz Intel Xeon processor with 1MB cache (parallel processing was turned off). A 3x7x7 3-D k-space kernel was chosen and a preprocessing step determined the source/target patterns and correlation values that needed to be generated. Figure 3 shows the 3 key computation steps of ARC: 1) all necessary correlation values are generated using the calibration data as input; 2) hybrid space (x, k_y, k_z) weights are generated for each source/target pattern from the correlation values; 3) missing data is synthesized in hybrid space, which has been shown to achieve additional computation time reductions (3). Once all missing data is synthesized, coil-by-coil images are reconstructed and combined using sum-of-squares.

Typical computation times for each step are noted in Fig. 3; computing all three steps can be completed in under a minute. Further optimization and parallelization should further reduce reconstruction time. Note that the first two steps can be initiated as soon as the calibration data is acquired (before the scan is completed).

Results Typical reconstruction results for the contrast-enhanced abdominal study are given in Fig. 4, showing that ARC is able to achieve good image quality, suppressing residual aliasing artifacts.

Discussion Correlation values speed up the calibration process and, coupled with hybrid space synthesis, enable ARC to efficiently reconstruct 2-D accelerated autocalibrating volumetric scans. Reusing correlation values across source/target patterns minimizes the penalty for variable-density acquisitions, which typically contain more patterns. This study demonstrates that ARC is able to achieve the image quality and reconstruction speed necessary to make it a clinically viable reconstruction method.

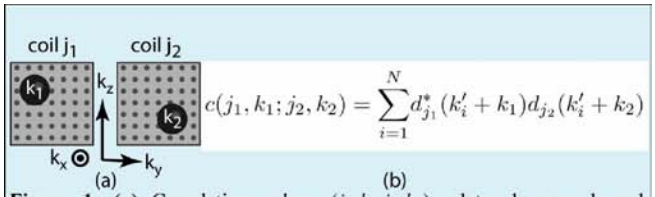

Figure 1: (a) Correlation value $c(j_1,k_1;j_2,k_2)$ relates k-space kernel location k_1 on coil j_1 to k-space kernel location k_2 on coil j_2. (b) For ARC, a correlation value can be written as the given sum, where N is the number of fit locations, k_i' is center of fit location i in the calibration region and $d_j(k)$ is the acquired datum on coil j at k-space location k. Every element of C_{ss} and C_{st} can be expressed as a correlation value.

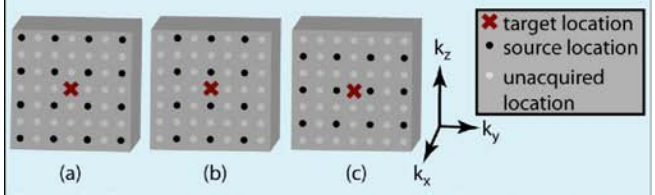

Figure 2: Three example source/target patterns. Shown in the (k_y,k_z) plane, the kernel size is 7x7. In the k_x direction a kernel width of 3 was chosen. Reconstruction weights dictate how data is synthesized at target locations from data at source locations on all coils.

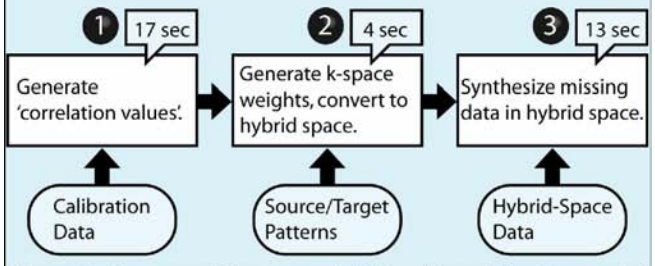

Figure 3: Key computation steps of ARC. Computation times noted correspond to the 2-D variable density acquisition described in the methods section. For this acquisition, step 2 generates weights for 32 source/target patterns. Since steps 1 and 3 comprise the bulk of the computation, the computational cost of multiple source/target patterns is small.

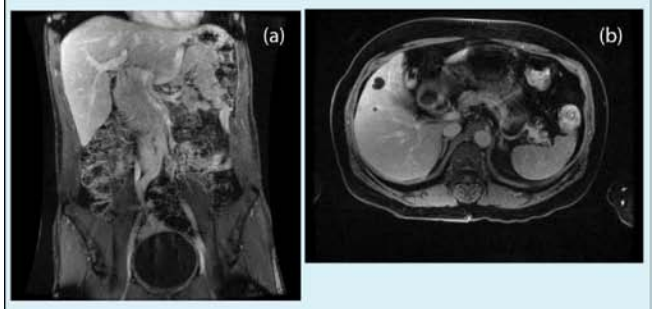

Figure 4: Example 2D-accelerated contrast-enhanced LAVA patient data with ARC reconstruction. (a) coronal plane (b) axial plane.

References

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(3) Brau et al. Proc ISMRM 2006, p2462.