Whole brain dynamic contrast enhanced imaging via compressed sensing techniques

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INTRODUCTION: 3D Dynamic Contrast-Enhanced (DCE) magnetic resonance imaging (MRI) is a well-established technique for noninvasive characterization of tissue physiology, and has found broad application in a wide range of diseases. Rapid T1-weighted acquisitions are used to record the uptake of an injected contrast agent. In the brain, recent developments in pharmacokinetic modeling methodology have enabled the separation of flow and permeability components in these T1-weighted acquisitions [1], providing a potential alternative to conventional T2*-weighted perfusion MRI and the limitations imposed by the use of EPI acquisitions [2]. Stringent sampling rate requirements are essential to acquire data rapidly enough to characterize the transit of the first pass of the bolus through the vasculature, resulting in corresponding limitations on spatial coverage and/or image resolution. In stroke, certain types of tumors, and other diseases of the brain in which lesions may be large or widely distributed, the ability to obtain rapid, high-resolution whole-brain coverage is highly desirable. In order to achieve this objective, acceleration factors of 3-4 relative to current state-of-the-art data acquisition are needed. Here, we demonstrate whole brain 3D DCE-MRI data, acquired at 2 mm isotropic voxel size and 4.3sec temporal resolution using a weighted pseudo-random undersampling scheme.

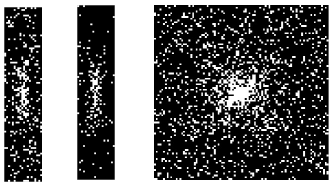

Figure 1: Three different sampling masks with acceleration rate R = 6 and R = 12 for simulation and R = 6 for real scan. The vertical direction is ky and the horizontal direction is kz. ky is 96 and kx for simulation is 20 and for real scan is 96.

METHODS: First a full dataset was acquired so that we could simulate different undersampling patterns and have a "truth" for comparison. A 3D spoiled gradient echo sequence was performed on a volunteer with a 1.5T whole-body MR scanner. A 0.1 mM/kg dose of Gd-BOPTA was injected. Scan parameters were TR = 2.84ms and TE = 1.8 ms, flip angle = 12 degrees and the acquisition matrix for this protocol is kx, ky, kz, t = 128 x 96 x 20 x 96. A 7/8 partial Fourier acquisition was used in the ky and kz directions, bandwidth = 490 Hz/pixel. In order to compare the difference between undersampled and fully sampled, the data from each coil was undersampled offline with a pseudo-random undersampling mask [3] with an acceleration rate R of 6 and 12, as shown in Figure 1.

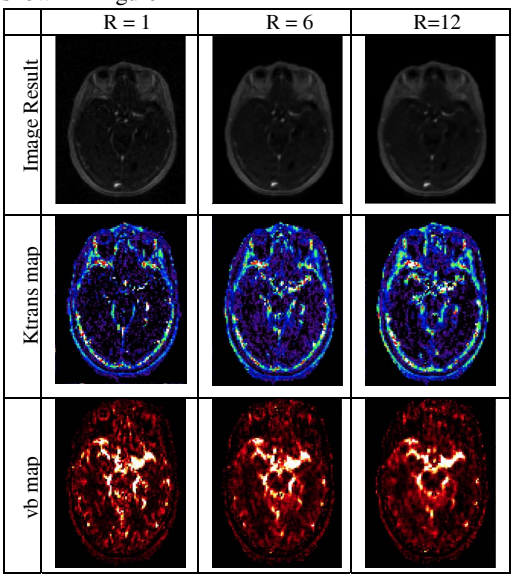

Figure 2: Comparison of fully sampled results and the undersampled simulation. The first row shows fully sampled data and undersampled reconstruction result with acceleration rates equal to 6 and 12 respectively. The second row is the Ktrans map for each of the results, and the third row is the v_b map.

To reconstruct the undersampled images, we used the STCR algorithm with total variation (TV) as the constraint term in both the temporal direction and the spatial directions [4].

Then a 3D spoiled gradient echo pulse sequence was modified to R=6 based on the variable density mask shown in figure 2. Imaging parameters for this full volume coverage "real" scan include: TR = 2.81ms and TE = 1.3ms, flip angle 12 degrees, and bandwidth = 490 Hz/pixel, which are the same as the fully sample protocol. Changes from the fully sampled case include the acquisition matrix kx, ky, kz, t = 128 x 96 x 96 x 96. With acceleration rate R = 6, there are 1536 phase encodes per time frame. This data was reconstructed in the same way as the fully sampled dataset described above.

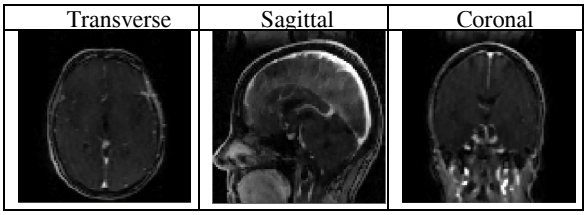

Figure 3: Three planes of the whole brain reconstruction image.

Both datasets were processed to generate pharmacokinetic parameter maps by regression of measured concentration-time curves [5] using a linearized Extended Tofts-Kety algorithm with both tissue permeability and blood volume terms [6]. Arterial input functions were determined separately for each reconstructed data set using a semi-automated algorithm for identifying and classifying the earliest enhancing voxels.

RESULTS: From the first "simulated" dataset with full sampling, it appears that R = 6 undersampling can give kinetic parameters comparable to the full data (Figure 2). For the "real" undersampled acquisition, 96 slice coverage of the whole brain was obtained, keeping the same temporal resolution and isotropic 2mm resolution. Figure 3 shows one slice from the full coverage data.

CONCLUSION: Accelerated DCE imaging of the brain can likely provide additional coverage, enabling isotropic resolution. Further studies are needed to assess the performance of the methods with respect to clinical tasks such as assessing the success of tumor treatment.

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