Feasibility of whole-brain dynamic contrast enhanced (DCE) MRI using 3D k-t PCA

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INTRODUCTION: Quantification of cerebral blood flow using dynamic contrast enhanced (DCE) MRI [1] has several advantages over conventional dynamic susceptibility contrast (DSC) MRI [2]. Most importantly, DCE-MRI avoids the inherent susceptibility artifacts of DSC-MRI, thus allowing for more reliable measurement of the arterial input function (AIF) and quantification of blood brain barrier leakage in brain tumors [3]. However, DCE-MRI currently suffers from low signal-to-noise ratio (SNR) and limited spatial coverage (in the order of 3-4 slices per second). The long acquisition times of DCE-MRI are due to interleaved saturation recovery (SR) preparation pulses, as well as incompatibility with echo planar imaging (EPI) which adds an undesirable degree of T2*-weighting. Theoretically, the SNR of DCE-MRI can be improved by acquiring data in 3D instead of in multiple 2D slices. Another advantage of 3D imaging is that the SR pulses can be skipped provided that the entire volume of interest is excited by each readout RF pulse. However, 3D DCE-MRI requires undersampling of k-space in order to maintain decent temporal resolution. In this study we present and investigate the feasibility of a non-SR-prepared 3D imaging sequence for whole-brain DCE-MRI. Image reconstruction from undersampled k-space data was performed using k-t PCA [3] and partial Fourier imaging, yielding a spatial coverage of 20 slices per second.

THEORY: For sufficiently short echo times the signal equation of the conventional SR-prepared DCE-MRI sequence is [4]:

$$SI = \Omega M_0 \sin(\alpha)\left((1-e^{-T_D R_1})(\cos(\alpha)e^{-T_R R_1})^{n-1} + (1-e^{-T_R R_1})\frac{1-(\cos(\alpha)e^{-T_R R_1})^{n-1}}{1-\cos(\alpha)e^{-T_R R_1}}\right),$$

where Ω is the receiver gain, M_0 is the fully relaxed (i.e., equilibrium) longitudinal magnetization, α is the readout flip angle, T_D is the delay between the SR pulse and the first α pulse, R_1 is the relaxation rate which depends on the contrast concentration and the baseline relaxation rate (i.e., before contrast injection), T_D is the repetition time, and n is the number of phase encoding steps (i.e., number of α pulses) before reaching the centre of k-space, defined as $k_y = 0$ for 2D imaging and $(k_y,k_z) = (0,0)$ for 3D imaging. Importantly, if n is sufficiently large the term $(\cos(\alpha)e^{-T_R R_1})^{n-1}$ vanishes, implying that SI no longer depends on T_D. In other words, DCE-MRI can be performed without the SR preparation pulse, provided that the number of phase encoding steps before reaching the centre of k-space is large enough. Strictly speaking this is true only if the α pulses are spatially non-selective, similar to the SR pulse. Otherwise inflow of unsaturated spins may cause a violation of the above signal equation in arterial blood. For 3D imaging the spatial extent of the α pulses is large, thus we assume in the following that the α pulses are spatially non-selective.

METHODS: Two patients with a brain tumor underwent DCE-MRI as part of their clinical routine examination. Imaging was performed on a 3.0 Tesla MR system (Achieva, Philips Healthcare) equipped with an eight-element head receive coil. One patient was examined using a non-SR-prepared 3D DCE-MRI sequence with a matrix size of 96×56×20 (k_x,k_y,k_z) and 40 time frames. We used a spoiled gradient echo readout, FOV = 230×182×160 mm^3, α = 15°, T_R = 3.9 ms, and T_E = 1.9 ms. For data undersampling we used 5-fold k-t PCA undersampling along k_y and k_z, and 62.5% partial Fourier imaging along k_y. The k-t PCA training data (matrix size = 96×11×11) were acquired interleaved with the undersampled data, resulting in a net acceleration factor of 6.1 and a temporal resolution of one 3D volume per second. In order for this sequence to satisfy the above equation, there were no delays between successive time frames, and the number of phase encoding steps before reaching the centre of k-space was n = 130. The second patient was examined using a conventional SR-prepared 2D DCE-MRI sequence [1] with fully sampled k-space data (matrix size = 96×61, slice thickness = 8 mm, FOV = 230×182 mm^3, α = 18°, T_R = 3.9 ms, T_E = 1.9 ms, and T_D = 120 ms). The fully sampled data were then decimated to simulate a 5-fold k-t PCA acceleration with 11 training profiles and 62.5% partial Fourier, allowing us to evaluate the loss in data quality due to undersampling.

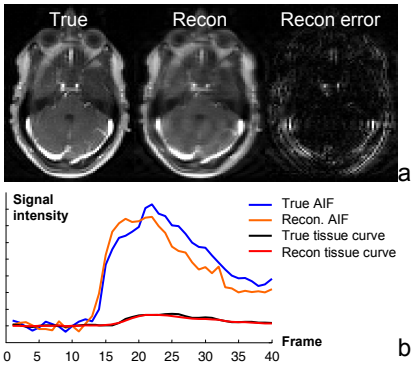

Fig 1. Comparison of fully sampled DCE-MRI data (True) and the simulated undersampled data (Recon) using the proposed data acceleration scheme. The Recon images are slightly blurred in both the spatial (a) and temporal (b) dimensions due to use of partial Fourier imaging. Notice, however, that this seems to affect only the AIF and not the tissue curve.

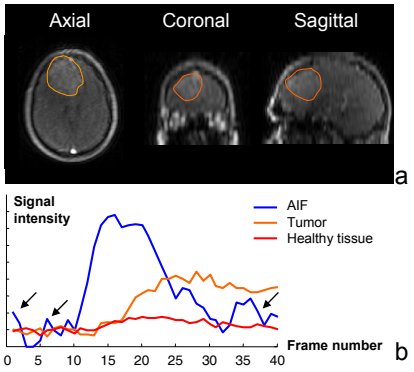

Fig 2. Whole-brain DCE-MRI in a patient with a large brain tumor (a) using the proposed data acceleration scheme. The signal intensity curve of the tumor shows strong contrast enhancement compared to healthy tissue (b), which is due to contrast leakage across the blood brain barrier. Notice that the inflow effects due to the lack of an SR pulse corrupt only the AIF (arrows) and not the tissue curves.

RESULTS: Figure 1 demonstrates the loss in data quality when using 5×k-t PCA and 62.5% half scan, as determined from simulations. The simulated undersampled images (denoted "Recon") appear slightly blurred along the phase encoding direction (Fig. 1a). This is a result of using partial Fourier imaging, which tends to blur small structures, such as vessels. The undersampled arterial input function (AIF) (Fig. 1b) measured in the right internal carotid artery shows temporal blurring. The loss in temporal fidelity with 5×k-t PCA is very quite small; hence, the temporal blurring of the AIF results from the spatial blurring introduced by partial Fourier, causing a contamination of true AIF signal with surrounding tissue signal due to partial volume effects. The tissue enhancement curves (Fig. 1b), however, do not suffer from temporal blurring. Figure 2 shows the results using the proposed non-SR-prepared 3D DCE-MRI sequence. The proposed sequence covers the entire brain (Fig. 2a), and the overall image quality and image contrast is similar to the conventional SR-prepared sequence. However, the AIF shown in Fig. 2b suffers from cardiac pulsation artifacts, suggesting that the spatial extent of the α-pulses is too small to avoid inflow effects. The tissue curves (Fig. 2c) show no significant cardiac pulsation artifacts.

CONCLUSIONS: Motivated by the limited spatial coverage of brain DCE-MRI, we have presented a new non-SR-prepared 3D DCE-MRI sequence that utilizes k-t PCA and partial Fourier imaging in order to simultaneously gain full spatial coverage and high temporal resolution. The proposed 3D sequence achieved a five-fold increase in spatial coverage compared with conventional 2D multi-slice approaches, while maintaining overall image quality and contrast. The use of partial Fourier introduced some spatial blurring, and the spatial extent of the α-pulses was insufficient to avoid inflow-related pulsation artifacts on the AIF. However, the tissue curves were not affected by these effects, suggesting that CBF quantification is indeed possible if the AIF is measured in a separate scan (i.e., using a dual-bolus approach) using a conventional SR-prepared sequence. We have already implemented the dual-bolus approach on our MR system and have scheduled more tumor patients to be examined in near future. This will allow us to perform a detailed comparison of whole-brain CBF maps with and without the dual-bolus approach.

REFERENCES 1) Larsson et al. JMRI 2008, 2) Ostergaard et al. MRM 1996, 3) Larsson et al. MRM 2009, 4) Pedersen et al. MRM 2009, 5) Larsson et al. MRM 2001