Initial Experience with Non-Contrast Enhanced Renal Angiography at 7.0 Tesla

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INTRODUCTION: The goal of this work was to develop the methods to perform non-contrast enhanced (non-CE) angiography studies of the renal vasculature at 7 Tesla. The potential advantage of performing these studies at 7T includes increased vessel conspicuity which would benefit a multitude of clinical applications including the evaluation of patients with renal insufficiency or failure, uncontrolled hypertension, and preoperative evaluation of renal vascular anatomy. The subset of patients with renal insufficiency or failure poses a unique clinical challenge in that their renal impairment may preclude the use of gadolinium-based contrast agents due to the possibility of developing nephrogenic systemic fibrosis, a debilitating and potentially fatal disease.

A common strategy to acquire non-CE renal angiography involves the use of an inversion magnetization preparation followed by a delay and data acquisition relying on in-flow enhancement of the vasculature. The inversion delay time (TI) is chosen to maximally suppress background signal from the parenchyma and venous blood. Performing these studies at 7T has the potential to improve the visualization of the first and second order branches as well as more distal arteries compared to lower field strengths and CE acquisitions (1). Along with the increased SNR expected at 7T, the higher T1s will result in longer inversion delays for background suppression and thus better in-flow enhancement. Previous studies at lower fields have shown similar techniques to compete with contrast-enhanced studies at identifying significant renal artery stenosis (2,3) and for evaluating the vasculature post transplantation (4). In this paper we address the technical and methodological challenges of performing these studies at 7T.

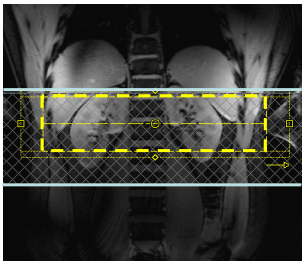

Figure 1:

METHODS: The two initial challenges that needed to be addressed were the inhomogeneous B0 and transmit B1 (B1+). While both of these are an increasing challenge for any target at ultra high field, the scale of this problem disproportionately increases as the target anatomy increases in size. To address B0 inhomogeneities, volumetric single breath-hold phase maps were acquired from which localized B0 shimming was performed [7]. Meanwhile, the complex transmit B1 field distributions were optimized for transmit homogeneity (as opposed to efficiency) based on a three slice, subject-dependent, single breath-hold, small flip angle calibration scan (5). A contoured region of interested was selected on each of the three calibration scan slices to target B1+ optimization to the region of the renal arteries. Every time a B1+ shim solution was applied, a quick (few seconds) small flip angle test image was obtained and compared against the pattern predicted by the B1 shim algorithm.

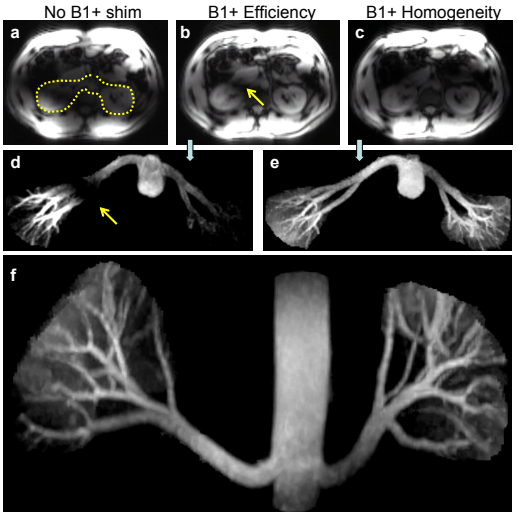

Figure 2:

The basic non-CE acquisition strategy was a respiratory triggered turbo-flash (TFL) which consisted of a slab selective inversion and chemically shift selective fat suppression followed by a gradient-echo readout. Planning of the axial imaging and inversion volumes are shown in Fig. 1 following the strategy presented by Liu et al. As the B1+ was optimized for homogeneity at the expense of efficiency, the maximum achievable B1+ made it difficult to achieve an inversion over the entire volume of interest. To improve inversion performance the standard adiabatic inversion pulse (HS1, 10 ms) was modified to a 25 ms HS4 which requires a lower peak B1+. Even when using this RF pulse, the maximum available RF power was required to suppress the background signal even on smaller individuals. Other basic imaging parameters included the following: 3.8 ms TR, 1.76 ms TE, 1000 ms TI, 8° nominal flip angle, GRAPPA = 2, 300 mm FOV, 72 slices with a nominal resolution of 1.0 x 0.9 x 1.0 mm^3. From the 3D source data, full MIPs were generated after manual cropping. The MRI system used for this study included a Magnex 7T, 90cm bore magnet with Siemens console and whole body gradients. An external 16 channel transceiver stripline array (6) was powered by a series of 16, 1 kW amplifiers (CPC, Pittsburgh, PA).

RESULTS: Figure 2 shows the results from a healthy subject. In Fig. 2 a, b and c, the middle of the three low flip angle test images are shown prior to any shimming (a) and after the application of a B1+ shim optimizing for transmit efficiency (b) and homogeneity (c). The region used for B1+ optimization is shown by the broken curve in figure 2a and was individually drawn on the other two slices of the calibration scan. The signal intensity in the low flip angle calibration scans has consistently been an excellent predictor on B1+ shim efficiency and homogeneity despite the large variation in body size and weight through the 5 volunteers included in the study. As expected the signal intensity in Fig. 2b shows the greatest B1+ in the kidneys but suffers from a local minimum across the right renal artery as shown by the yellow arrow and as demonstrated in the axial MIP, Fig. 2d. While the homogenous solution, Fig. 2c, results in a loss in transmit efficiency, the local minimum present in the efficient solution is absent and enables the acquisition of the complete vasculature as demonstrated by the axial MIP in Fig. 2e. A coronal MIP of the same acquisition with the homogeneous solution is shown in Fig. 2f.

DISCUSSION: In this preliminary study, we have found that the targeted homogeneous B1+ shimming solution based on the small flip calibration scans was necessary for consistently obtaining both branches of the renal arteries. In our hands, magnitude B1+ mapping in the large flip angle regime was a less robust approach with high sensitivity to physiological motion whatever technique was used, including AFI and MP-TFL, therefore was not an option for optimizing B1+. Due to the robust and rapid nature of the single breath hold small flip angle calibration scan, it is directly suitable as-is for use in clinical studies.

In order to achieve the results demonstrated in Fig. 2, the power to the inversion pulse had to be manually set to the maximum, and yet, full inversion was still not obtained. This can be observed in Fig. 2f where the parenchyma of the kidneys is not fully suppressed. Even without complete suppression the there is the vasculature past the second order branches can be appreciated. When optimizing for efficiency, as in Fig 2d, background suppression was incomplete and also experienced the loss of critical vascular anatomy. One parameter which was not explicitly optimized in these studies was the excitation flip angle. After careful analysis of our data it was evident that the small flip angle used for excitation was in some cases too high. In figure 2d, an excitation flip angle was used resulting in high SNR in the right kidney vessels while saturating the spins in the left. The situation is better in the homogeneous case. While there is still some shading present in the main branch of the left kidney in Fig. 2e and 2f, the excitation angle was decreased thus maximizing in-flow enhancement throughout.

This initial study has highlighted several areas for future development. First, the acquisition requires only one inversion RF pulse every respiratory cycle followed by a series of low flip angles over a short acquisition window. Therefore, these studies are limited only by peak B1+ and not SAR. Any improvement in coil or transmit chain efficiency would benefit these studies. In addition, RF pulse design could play an equally important role as there are several methods available to obtain adiabatic inversion with lower peak B1+ requirements. Further optimization of the inversion delay and excitation pulse will also improve the quality of the final results by maximizing both CNR and SNR. Finally, once optimized, the goal would be to compare these acquisitions against the best non-CE SSFP sequences currently being investigated at lower field strengths with respect to the identification of stenosis and overall evaluation of the renal arteries.

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