Influence of contrast-dependent T2* effects on DCE-MRI of the prostate at 7T

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INTRODUCTION: The desire to perform DCE-MRI studies at ever increasing magnetic fields comes from the promise of increased spatial and temporal resolution. However, to effectively use DCE-MRI at high fields, the increasing influence of T2* on the typically T1-weighted acquisition must be understood to avoid misinterpretation of results. While R1 relaxivity has been shown to slightly decrease [6], R2* relaxivity increases with field strength and can dominate signal changes especially at higher contrast agent concentrations. In addition, the strong effect of compartmentalization on T2* shortening can also change the shape of the contrast enhancement curve and thus the quantification of pharmacokinetic parameters. The goal of this paper was to investigate the influence of T2* shortening on contrast enhancement curves at 7T in the prostate and to determine the impact on pharmacokinetic modeling.

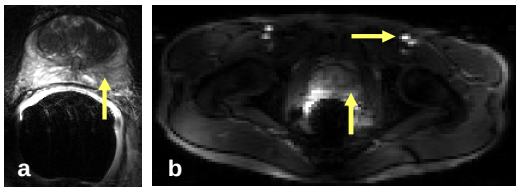

Figure 1: a) T2w anatomical and b) DCE axial views.

METHODS: 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 stripline array was used for RF transmission [1] of which 15 channels were used for receive in combination with a receive-only endorectal (ERC) coil. The receive-only ERC coil was similar to the transceiver coil previously presented [5] with the addition of both active and passive decoupling. For transmit, a series of 16, 1 kW amplifiers (CPC, Pittsburgh, PA) were optimized for transmit efficiency in the region of the prostate [2] while power monitoring allowed FDA guidelines on local SAR to be followed. Limits on power were based on previous FDTD modeling studies [5].

Imaging data were collected on a healthy volunteer (GFR > 90) under an IRB-approved protocol. Initially, multi-slice axial and sagittal T2w anatomic TSE images were acquired for visualization of anatomy and planning. For DCE-MRI, a single dose (0.1 mmol/kg) of Gd-DTPA (Magnevist, Berlex Laboratories) was delivered by hand injection at a rate of approximately 3 cc/sec followed by a 20 ml saline flush at the same rate. The DCE-MRI data was obtained with a 3D gradient echo sequence with the following parameters: 350 mm FOV, 2.7x2.7x3 mm^3 resolution, 9.1 ms TR, 20° nominal flip angle, 24 slices, 128 kHz BW and 30 dynamics with a temporal resolution of 15 sec for a total acquisition time of 7.4 min. The RF power was calibrated to achieve the nominal flip angle in the prostate. Multiple unipolar readouts were acquired during each TR at echo times of 3.1, 5.1 and 7.1 ms to allow the calculation of pixel-wise T2* values throughout the entire contrast enhancement curve. T2* values were calculated fitting the signal intensity of the 3 echoes to the equation A0*exp(-TE/T2*) where A0 is a combination of M0 and noise. Pharmacokinetic parameters were calculated by fitting the tissue signal intensity-times curves with a two compartment general kinetic model (GKM) using an arterial input function (AIF) obtained from the region of the femoral artery. The GKM model yielded the parameters K^{Trans} (forward volume transfer constant, min^{-1}), k_{ep} (reflux rate between the extracellular space and the plasma, min^{-1}) and the ratio of the two providing the fractional extravascular-extracellular space ($v_e = K^{Trans}/k_{ep}$). These parameters were calculated both before and after correcting for time dependent T2* values. All analyses were carried out in IDL (ITT, Boulder, CO).

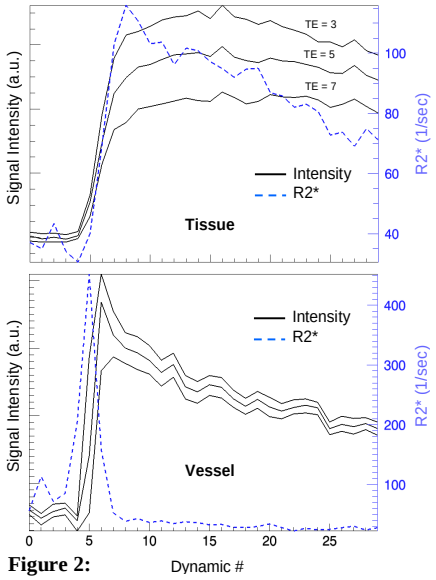

Figure 2:

RESULTS: Figure 1a shows the anatomic T2w image at the level of the analyzed DCE slice, which is shown at dynamic #7 after contrast arrival in the prostate (Figure 1b). The vertical arrow shows the location of the tissue curve while the horizontal arrow shows the location of the vessel curve. The signal intensity and R2* (1/T2*) versus dynamic curves are shown for tissue (upper plot) and vessel (lower plot) in Figure 2. Figure 3 shows the original (TE=3.1 ms) and T2*-corrected enhancement curves for both the arterial and tissue contrast enhancement curves. The corrected curves were calculated by removing the static and time dependent T2* signal loss resulting from a combination of global B0 inhomogeneities, intrinsic local variations in susceptibility and contrast passage. Pharmacokinetic modeling using the original enhancement curves from the shortest echo time data (3.1 ms) resulted in the following values: $K^{Trans} = 0.33$, $k_{ep} = 1.00$, and $v_e = 0.33$. Fitting the original tissue enhancement curve with the T2* corrected AIF resulted in the values: $K^{Trans} = 0.19$, $k_{ep} = 0.60$, and $v_e = 0.32$. Fitting the T2* corrected tissue enhancement curve with the corrected AIF resulted in the values: $K^{Trans} = 0.28$, $k_{ep} = 0.70$, and $v_e = 0.40$.

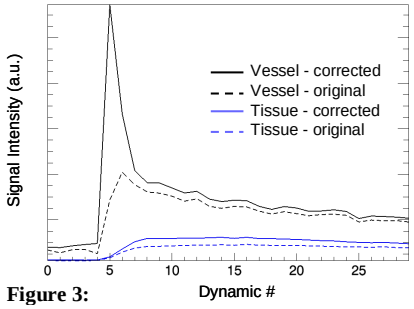

Figure 3:

DISCUSSION: Unlike previous studies at 3T where the AIF (or vessel signal) alone was corrected for T2* effects [3], at 7T there is also a tremendous influence of T2* on the tissue contrast enhancement curve which requires correction. This is not only evident in figures 2 and 3 but is most importantly demonstrated by the variations in the fitted pharmacokinetic parameters. Correcting for T2* of the AIF alone reduced the values of K^{Trans} and k_{ep} by approximately 40% while their ratio, as reported by v_e, remained about the same. After correcting both the vessel and tissue curves, K^{Trans} increased relative to k_{ep} resulting in a 25% higher v_e.

There are several factors which contribute to the dramatic signal changes observed during the bolus passage of contrast in both tissue and vessel which confound the linear relationship between T1-weighted signal intensity and Gd-DTPA concentration [Gd]. It is well understood that R2* increases rapidly in parallel to increased compartmentalization [4]. Upon initial contrast arrival at the tissue, it is most compartmentalized as it resides in the microvasculature and it has not had a chance to diffuse throughout the extracellular-extravascular space (v_e). As contrast leaks from the vasculature and diffuses throughout the tissue, compartmentalization decreases while [Gd] can increase. This is observed in the upper plot of Figure 2, where a precipitous drop in tissue R2* after dynamic #9 is mirrored by increasing T1w signal intensity (TE = 3 ms curve) out to dynamic 15. As the short echo time T1w signal intensity is closely related to [Gd] we can assume that local levels of contrast are still increasing while R2* drops. The T2* corrected signal intensity curve also increases over this same period but is difficult to appreciate because of its scale in Figure 3.

There are multiple ways to reduce or eliminate the T2* effects in DCR-MRI studies including: 1) lowering injection rates and/or reducing dose, 2) collecting multi-echo data for correction purposes and 3) acquiring data with ultra-short TE sequences. While multiple echo data was acquired at the expense of temporal resolution, it is feasible that such sacrifices are not necessary because of the potential of high parallel imaging acceleration factors possible with the multi-channel coils being used for high field body imaging [1]. In addition, above and beyond correcting the contrast enhancement curves, the calculation of time-dependent T2* during contrast passage may in itself reveal useful information about the tissue microenvironment; an area which requires further investigation.

REFERENCES: [1] Snyder et al. Proc Intl Soc MRM 2007; 164. [2] Metzger et al. MRM 2008;59(2):396. [3] de Bazelaire et al. Eur Radiol 2006;16: 2083. [4] Kjolby et al., MRM 2006;56: 187. [5] Metzger et al. Proc Intl Soc MRM 2008;16:171. [6] Vander Elst et al. Invest Radiol 1988; 33(11): 828.

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