

Comparison of DCE-MRI and DCE-CT in patients with renal cell carcinoma: effects of temporal resolution and total measurement time

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INTRODUCTION: Temporal resolution is crucial for accurate fitting of dynamic contrast-enhanced MRI (DCE-MRI) data. In particular, accurate determination of the AIF peak facilitates the fit of the volume transfer constant K_{trans} . The volume fraction of the extravascular extracellular space (EES), v_e , is determined during later stages of uptake. Dynamic contrast-enhanced computed tomography (DCE-CT) is a growing method of assessing tumour perfusion. It offers improved temporal resolution over MRI, but the number of images that can be acquired is limited by radiation dose. Although their mechanism of signal generation is different, both MRI and CT contrast agents (CAs) diffuse from the vasculature to the EES and can be analyzed by the same pharmacokinetic model. In this study, data from patients who have undergone DCE-MRI and DCE-CT are fit to a two-compartment Kety-Tofts model². The effect of temporal resolution and total measurement time on the fit is examined.

METHODS: Eight same-day scans of patients with renal cell carcinoma were performed by MRI and CT. DCE-MRI data were acquired at 1.5 T (GE Signa, Milwaukee, WI) using a multiphase 3D LAVA (SPGR-based) sequence (TR=3.173 ms, TE=0.968 ms, 0.75 NEX, 128² matrix, $\alpha=15^\circ$, FOV=38-42 x 38-42 cm²). Twelve coronal slices, 8 mm thick, were acquired every 3.7 s for ~5 minutes. Twenty seconds into the scan, 0.2 mmol/kg Gd-DTPA-BMA (Omniscan, GE Healthcare) was injected. Pre-contrast T1s were determined from 2D axial fSPGR scans with $\alpha=15^\circ$ and 30° (TR = 50 ms, 128² matrix, 15 x 10 mm slices, FOV = 38 x 38 cm²) and used for normalization. CT data (GE LightSpeed VCT) were acquired every second for 55 s, followed by one scan every 4.3 s for 52 s. Visipaque (1 mL/kg, GE Healthcare) was injected 5 s into the scan. Slice thickness was 5 mm, with resolution from 0.5-0.7 mm² in plane.

For MRI, 3-4 regions of interest (ROIs) in different slices were drawn around the tumour mass by a radiologist. For CT, 7-8 ROIs were drawn around the volume, except in one patient where volume limited this to five slices. ROIs were also drawn in the aorta to obtain an AIF. Data from each tumour ROI were then fitted to the Kety-Tofts model², which describes CA movement between the blood plasma and extravascular space by: $C_e(t) = K_{trans} / v_e \int C_p(\tau) e^{K_{trans}(t-\tau)/v_e} d\tau$ [Eq. 1], where C_p and C_e are concentrations of contrast agent in the plasma and EES, K_{trans} is the volume transfer constant from the plasma to the EES and v_e is the EES volume fraction. In addition, translation of C_p in time was allowed to account for the delay in bolus arrival between the aorta and tumour ROIs. The attenuation in the CT data was assumed to be linearly related to CA concentration³. For MRI, the signal was related, via the T1 relaxation time, to the CA concentration by assuming fast intra- to extracellular water exchange. For both imaging modalities, the change in signal due to blood pool CA was assumed to be negligible. A haematocrit of 0.4 was assumed. Fit parameters from each ROI were then averaged across all slices to give a value for the whole tumour. Errors in each fit parameter were calculated by adjusting that parameter and re-fitting the remaining parameters until the reduced χ^2 distribution differed from the χ^2 of the original fit with 68% confidence⁴. To examine the effect of temporal resolution, Δt , on parameter values and fit errors, the CT data (AIF and tumour) were refit to Eq. 1 using only every third data point for the first 55 s, plus the data from the final 52 s, creating a temporal resolution comparable to that of the MR data. To examine the effect of the total measurement time, T, the MRI data were re-fit using only the first two minutes of data, comparable to the total CT measurement time.

RESULTS: Samples of the uptake curves and AIFs are shown in Figure 1 for (a) MRI and (b) CT. Figure 2 compares the fit parameters (a) K_{trans} and (b) v_e between CT and MRI. Error bars indicate the average error in the fitting procedure, not the standard deviation across ROIs. The fit of the MRI data for shortened T (Figure 3a) shows systematic underestimation in v_e (except one case) by an average of 16% and a fit error increase of 103%. K_{trans} values (not shown) also increase 27%, with a fit error increase of 66%. CT data for $\Delta t \sim 3$ s (Figure 3b) show that K_{trans} is underestimated by an average of 16% at lower temporal resolution and the average fit error increases by 60%. There is also a much smaller 4% increase in v_e (data not shown).

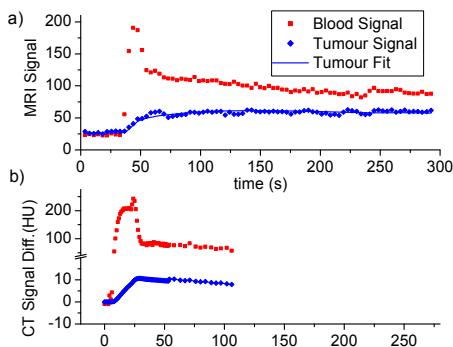


Figure 1 Uptake curves and arterial input functions for (a) MRI and (b) CT. Note that the individual ROIs do not correspond to one another. The solid line shows the fit to Eq. 1

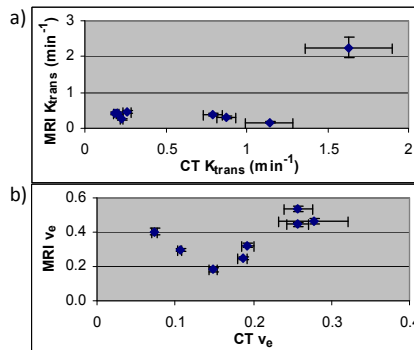


Figure 2 Comparison of fit parameters between CT (x-axis) and MRI (y-axis). (a) K_{trans} shows no correlation. (b) v_e is weakly correlated with one outlier. Error bars show the average fit error.

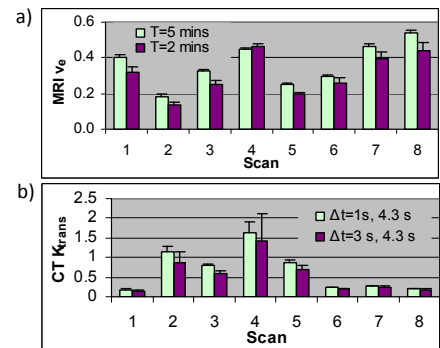


Figure 3 The original values of the fit parameters are shown in green. The purple bars show how the fit changes (a) for the MRI parameter v_e when T is shortened and (b) for the CT parameter K_{trans} when the Δt becomes longer.

DISCUSSION: The value of K_{trans} from CT did not correlate with that from MRI. This is somewhat surprising. Although the different sizes of MRI and CT contrast agents may exhibit different tracer dynamics, the lack of correlation indicates vastly different mechanisms of perfusion. There was a weak correlation between the v_e values for six of the seven data points. The fast exchange assumption used to fit the MRI data was not valid at high concentrations of gadolinium where water relaxes much more quickly than it exchanges between the intra- and extracellular spaces. Because CT signal is determined directly by the contrast agent, it was not subject to such errors.

As previously demonstrated, high temporal resolution was necessary to determine the peak of the AIF and fit K_{trans} accurately. The underestimation and increased fit errors observed at larger Δt values for CT were most pronounced for large values of K_{trans} where uptake occurred rapidly and was most dependent on early time points of the AIF. Similarly, v_e was determined largely by the washout phase, so limiting time of MRI data acquisition caused an underestimation of v_e and an increase in the fit error. However, because the K_{trans} and v_e parameters are coupled during the washout phase, there is also a small, compensating overestimation in K_{trans} . The higher temporal resolution of DCE-CT may therefore lead to improved estimation of K_{trans} , particularly when K_{trans} is high, but radiation dose limits the number of scans acquired during the washout phase, which can underestimate the volume into which the tracer leaks and lead to increased errors in the fitting of both K_{trans} and v_e .

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