

# Optimisation of Time-resolved angiography With Stochastic Trajectories (TWIST) for Dynamic Contrast-Enhanced MRI in Head and Neck Cancer

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**Introduction:** View-sharing techniques allowing for significant reduction of acquisition times are suited for Dynamic Contrast-Enhanced (DCE) MRI measurements where high temporal resolution is required<sup>1</sup>. However undersampling of peripheral parts of k-space may lead to impaired characterization of enhancement curves and ringing artifacts<sup>2</sup>. In this study we investigate the effects of different k-space undersampling patterns with Time-resolved angiography With Stochastic Trajectories (TWIST) on pharmacokinetic (PK) DCE vascular parameters.

**Materials and methods:** DCE images acquired prior to treatment in a group of patients (n=8) with stage III and IV squamous cell carcinoma of the head and neck (SCCHN) were used in the study. A representative slice with the maximum lesion diameter was identified and a set of images measured at each DCE time point was Fourier transformed to obtain k-space data using Matlab (Mathworks, Cambridge MA). K-space was divided into central and peripheral parts and 5 different sizes of the central part ( $A = 2, 10, 20, 33, 50\%$ ) were considered. In the next step the peripheral part was undersampled, such that a defined percentage of k-space points ( $B = 10, 33, 50\%$ ) was randomly sampled. For each image all points in compartment A and a percentage of the points in compartment B are used, with missing portions of B copied from adjacent time points. Constructed k-spaces were Fourier transformed to obtain a set of simulated TWIST images used for DCE pharmacokinetic calculations. This approach is similar to the one used by Song et al. in a renography phantom study<sup>2</sup>.

MRI was performed at 1.5T (Philips Intera, Philips Medical Systems, Best, Netherlands). The DCE protocol included a transaxial 3D FFE sequence ( $TE = 1$  ms,  $TR = 4$  ms,  $FOV = 256 \times 256$  mm<sup>2</sup>, 10 slices,  $2 \times 2 \times 5$  mm<sup>3</sup> voxel, 1.5 s temporal resolution) for acquisition of proton density-weighted images ( $FA = 4^\circ$ ) followed by 100 dynamic acquisitions ( $FA = 20^\circ$ ) during which gadolinium contrast was injected (0.2 mg/kg, Dotarem®, Guerbet, France).

DCE PK modelling was performed using the software package MRIW (Institute of Cancer Research, UK)<sup>3</sup> with the extended Kety model<sup>4</sup>. A set of parameters was derived including: Ktrans - volume transfer constant between blood plasma and extracellular extravascular space (EES), Ve - total EES volume and Vp - total blood plasma volume. DCE maps were produced for each parameter and lesion ROI (Fig. 1, left). Percentage differences between reference and TWIST simulated DCE maps were calculated on a pixel-by-pixel basis (Fig 1, right). The ROI median of absolute percentage differences was calculated for all tested A/B combinations and patients. The results averaged for all patients were presented in a form of a parameter maps (Fig. 2&3) and a standard deviation map was presented for Ktrans differences as a measure of intra-patient variability.

**Results:** Median values of Ktrans, Ve and Vp parameters calculated for lesion ROIs were  $0.181 \text{ min}^{-1}$ ,  $0.202 \text{ ml}$  and  $0.013 \text{ ml}$  respectively. DCE parameter inaccuracies (Fig 2 & 3) depend on the choice of undersampling pattern and vary between DCE parameter types. The Ktrans and Ve differences are under 10% for  $A > 30\%$  and  $B > 20\%$ . The Vp was most affected by TWIST undersampling and differences lower than 10% can be achieved for  $A > 35\%$  with an extended region between  $B: 25 - 35$  and  $A > 25\%$ .

**Discussion:** We demonstrated the influence of the TWIST view-sharing technique on DCE calculations using heterogeneous groups of primary and nodal tumor sites, taking account of the relatively wide range of underlying perfusion regimes observed in SCCHN. The high temporal resolution of the conventional 3D FFE sequence allowed a reliable calculation of DCE parameters, however measurements were affected by partial volume effects and limited sup.-inf. site coverage. This is especially problematic in the case of longitudinal DCE measurements during the course of a treatment, where the decreasing volume of responding lesions approaches the slice thickness and the number of viable voxels is limited. The use of TWIST can overcome these limitations providing a good compromise between coverage, temporal and spatial resolution.

**Conclusion:** The TWIST sequence can reliably be used for pharmacokinetic DCE MRI using a range of undersampling patterns. The parameter TWIST maps created in the study can be used to balance temporal and spatial resolution demands with the quality of enhancement curve characterisation.

**References:** <sup>1</sup>Davenport MS. et al. J. Magn. Reson. Imaging. 2013;38(2):329-35, <sup>2</sup>Song T. et al. Magn. Reson. Med. 2009; 61(5):1242-1248, <sup>3</sup>D' Arcy J.A. et al. Radiographics. 2006;26(2):621-32, <sup>4</sup>Tofts P.S. et al. J. Magn. Reson. Imaging. 1999;10(3):223-232.

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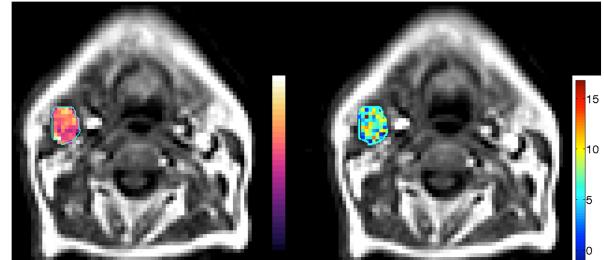


Fig. 1. Example of nodal lesion Ktrans map (left) and absolute percentage Ktrans difference for TWIST (right) using  $A=33\%$  and  $B=33\%$  sampling. Ktrans median =  $0.367 \text{ min}^{-1}$  (range: 0 –  $0.58 \text{ min}^{-1}$ ).

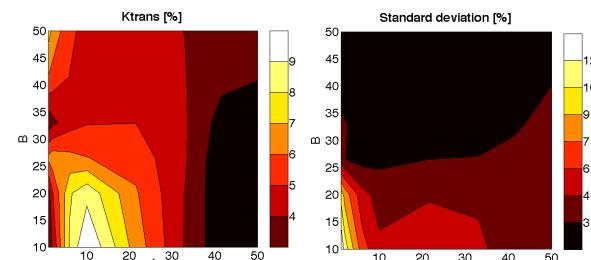


Fig. 2. TWIST Ktrans parameter map (left): an absolute value of median percentage difference (n=8) of Ktrans calculated using reference (no TWIST) and simulated TWIST DCE image sets. Standard deviations for the percentage difference are shown on the right. Simulated TWIST undersampling partition sizes:  $A = 2, 10, 20, 33, 50\%$  and  $B = 10, 33, 50\%$ .

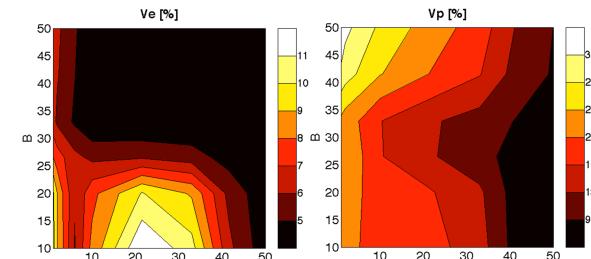


Fig. 3. TWIST Ve (left) and Vp (right) parameter maps (n=8).