Correction of B_1-field inhomogeneity of surface coils in the analysis of DCE-MRI experiments

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Introduction Dynamic contrast-enhanced MRI provides quantitative information on the transfer constant (K^{trans}) of tumour tissue, and has a variety of applications in tumour grading and diagnosis and as an endpoint in drug studies, particularly for antivascular or antiangiogenic agents. For subcutaneous animal tumour studies, surface coils provide a useful enhancement in SNR. However, if a surface coil is used both for excitation and reception and a uniform flip angle is assumed in the T_1-weighted dynamic series, substantial errors may be introduced into the determination of contrast agent concentration [Gd], leading to errors in the dynamic parameters. These errors can be avoided when the spatial flip-angle distribution is known and taken into account during the analysis of the images (Figure 1).

Methods MRI was performed on a Varian Unity Inova spectrometer with a 4.7T horizontal bore magnet of inner diameter 154mm. Data were acquired from a RIF-1 tumor grown subcutaneously in the flank of a C3H-mouse. The tumor hung freely in a short 2.5 turn solenoid coil of inner diameter 1.75cm. A pre-contrast T_1 map was acquired using IR FLASH [1]. T_1-weighted gradient-echo (nominal flip-angle 50°, TR=40, 60, 80, 100, 120, 160, 200, 400, 1000, 5000 ms) images were then acquired. Finally, DCE-MRI was performed using T_1W GE, nominal flip-angle 50°, TR=80ms, 128x128 points, time resolution 10.24 seconds. A total of 125 images were acquired and 0.3mmol/kg GdDTPA was injected manually after the 5th scan. A T1 map calculated from the IR-images was used together with the gradient-echo images to calculate maps of the spatial flip-angle distribution. To reduce noise related errors, a simulated flip-angle map was calculated using a three-dimensional model of the coil and the Biot-Savart law, and its amplitude and spatial position fitted to the experimental flip-angle maps. DCE-MRI data were analyzed by the model of Tofts and Kermode, using the fitted flip-angle for each voxel. This resulted in K^{trans} and v_e values unbiased by the spatial flip-angle variation of the surface coil (Figure 1).

Results and Discussion As shown earlier in a phantom study [2], errors in the estimated change in R_1 can be corrected within limits dictated by flip angle variation and SNR. Here we show the applicability of this method to an *in-vivo* DCE-MRI experiment. The non-uniform flip-angle across the image affects both the intensity as well as the T_1-weighting in the series of T_1-weighted gradient echo images. Using the correct flip-angle value for each voxel results in parameter maps (e.g. K^{trans}) that are not affected by the inhomogeneity of the B_1-field (Figure 1). Figure 2 shows a histogram of K^{trans} values of five slices through the tumor calculated assuming a uniform flip-angle (black) and using the simulated flip-angle maps (gray). There is a substantial shift in the histogram overall. Close to the centre of the coil, the K^{trans} values are similar. In regions where the flip-angle is reduced, the difference is much larger; over the region of interest ringed in Figure 1, the mean estimated K^{trans} values were 5.71×10^{-5} [s^{-1}] (assuming a uniform flip-angle of 50°) and 1.54×10^{-4} [s^{-1}] (using the flip-angle map). Without flip-angle correction, this sizeable error could be misinterpreted as a true difference between tumour rim and core.

In contrast to other methods [3, 4] this method is not based on the use of uniform phantoms. Those methods inherently assume that the field generated by the coil does not change between the phantom and the object under investigation. Methods which do not use a theoretical model for the magnetic field of the coil might carry over noise from the acquired images to the estimated parameters, which is not desirable. Another method that uses a simulated model of the reception profile using the Biot-Savart law needs MR-visible reference markers [5]. The method presented here can be used in two ways. Either the experimental flip-angle map can be used directly (in which case some kind of denoising might be advantageous) or the simulated flip-angle map can be used. The latter procedure has the advantage of not introducing artificial noise in the estimated parameters via a noisy flip-angle map. In any case when using a multi-slice acquisition, no reference markers are necessary, which makes this method easy to implement for data acquisition as well as for the data analysis. An alternative solution would use a volume coil to excite the signal and a surface coil for reception, eliminating flip-angle variation. This has the added advantage of increasing the signal received from tissue remote from the coil, enhancing the SNR, though not all small-animal systems readily allow this scheme. The original rationale for this work was to improve the performance of a DCE-MRI method using interleaved tumour/tail vein acquisition to provide an AIF with each acquisition [6,7]. This required switching between a tumour surface coil and a small solenoid surrounding the tail, which precluded the use of a volume-excite/surface receive arrangement as our system cannot switch three coils simultaneously. The combination of the excellent sensitivity of the surface coil (Figure 2) and the flip angle correction results in a dramatic reduction in both noise-related and spatially varying systematic errors in determination of DCE-MRI parameters.

Acknowledgements This work was supported by Cancer Research UK [CRC] grant C2/A1209

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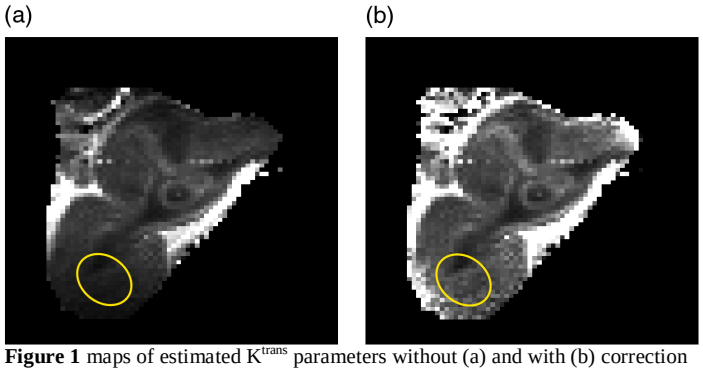

Figure 1 maps of estimated K^{trans} parameters without (a) and with (b) correction for B_1-inhomogeneity.

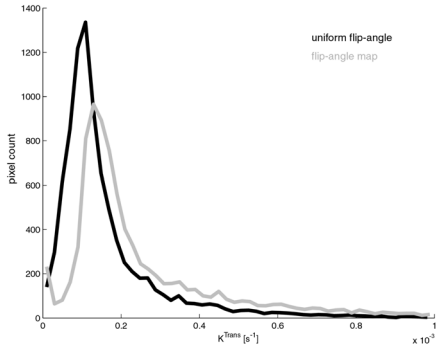

Figure 2 Histograms of K^{trans} values without (black) and with (gray) correction for B_1-inhomogeneity