Comparison of the Standard Gadolinium Concentration and Signal Difference Methodologies for Computation of Perfusion Parameters in DCE-MRI at Various SNRs

H. K. Song[1], Y. Xue[1], J. Yu[1], S. Englander[1], H. C. Kang[1], and M. A. Rosen[1]
[1]Department of Radiology, University of Pennsylvania, Philadelphia, PA, United States

Introduction: In standard DCE-MRI, T_1-weighted images are acquired rapidly during gadolinium contrast injection, and the concentration [Gd] in tissue is estimated based on the measured T_1s, from which the tumor perfusion parameters can be computed. Since the time-intensity curve first needs to be normalized to the pre-contrast signal, noisy baselines can lead to measurement biases. Recently, methods to compute perfusion without the need to first compute T_1 or gadolinium concentration have been reported (1-3). The technique is based on the observation that the signal difference, after pre-contrast value is subtracted, is approximately proportional to [Gd]. Since normalization (division) by the baseline signal is unnecessary, the latter method for computing perfusion may be more robust in the presence of increased noise levels. The goal of this work is to systematically investigate the performance of the signal difference technique and compare it to the conventional method for computing perfusion under various image SNRs.

Methods: The signal difference ΔS (after subtracting the pre-contrast baseline) for a spoiled gradient echo sequence can be approximated as (2):

$$\Delta S \propto a \cdot \sin\alpha \cdot TR \cdot \left[\frac{R_1}{1-\cos\alpha \cdot (1 - TR \cdot R_1)} - \frac{R_{10}}{1-\cos\alpha \cdot (1 - TR \cdot R_{10})} \right] \quad [1]$$

where a is a factor that includes coil sensitivity, R_1 and R_{10} are the time-dependent and native longitudinal relaxation rates, respectively, and α is the flip angle, and the condition $TR << 1/R_1$ is assumed. Under the latter assumption, ΔS can also be shown to be proportional to [Gd]:

$$\Delta S \propto a \cdot \frac{\sin\alpha}{1-\cos\alpha} \cdot TR \cdot \Delta R_1 = a \cdot \frac{\sin\alpha}{1-\cos\alpha} \cdot TR \cdot r \cdot [Gd] \quad [2]$$

where $\Delta R_1 = R_1 - R_{10}$, and r is relaxivity. Since ΔS is approximately proportional to [Gd], ΔS could be used directly in the first order kinetics model without the need to compute [Gd].

Monte Carlo simulations were performed to assess the performance of the signal difference method and compare the results to conventional concentration-based measurements of perfusion parameters. For the arterial input function, an experimentally-derived functional form based on a population-averaged input function described previously (4) was utilized. Two pairs of [K^{trans}, v_e] values were used to generate the tumor signal ([0.4 min^{-1}, 0.4] and [1.2 min^{-1}, 0.4]) and varying amounts of normally distributed zero-mean complex noise were subsequently added to the tumor and AIF signal. Sufficiently high temporal resolution of 1.7 sec per frame was assumed (5), with TR=3.2ms. The following parameters were also used in the simulation: $T_{10\ blood}$=1200ms and $T_{10\ tumor}$ = 800ms, relaxivity r_1 = 4.0 mM^{-1} sec^{-1}. For the AIF, 25-pixel averaging of the magnitude blood signal was performed, while the tumor signal was processed on a pixel-wise bases (no averaging).

Results and Discussion: Figure 1 shows the results of our simulations for K^{trans}=0.4 min^{-1}. At a flip angle of 25°, which is on the order of what is typically used for DCE-MRI, the accuracy of K^{trans} for the conventional [Gd] method is superior except for the lowest SNRs. Because the [Gd]-ΔS relationship deviates more significantly from linearity at smaller flip angles, there is a bias in computed K^{trans} values for the signal difference method, resulting in substantial errors (approx. 20%) even at higher SNRs. However, at a higher flip angle of 50°, the [Gd]-ΔS relationship becomes more linear (as the denominators in Eq.1 become even less dependent on R_1 and R_{10}), and K^{trans} values become significantly more accurate. At either flip angle, v_e is much more accurate with the signal difference method, although the results are more favorable at the lower angle. Similar results were found for K^{trans}=1.2, though the errors were slightly higher.

Several assumptions were made in our simulations. For the standard concentration-based method, it was assumed that intrinsic lesion T_1 is known and that the nominal flip angle was accurately applied at the locations of the AIF and tumor. Errors in these parameters will likely further increase the measurement errors. For the signal difference method, since absolute signal intensity is directly used in the calculations, receiver coil sensitivity was assumed to have been determined, e.g. by taking the ratio of receive coil and body coil images. Although the latter method does not require precise knowledge of the flip angle, it does require that the flip angle is similar at the locations of the AIF and tumor, and such was assumed in our work.

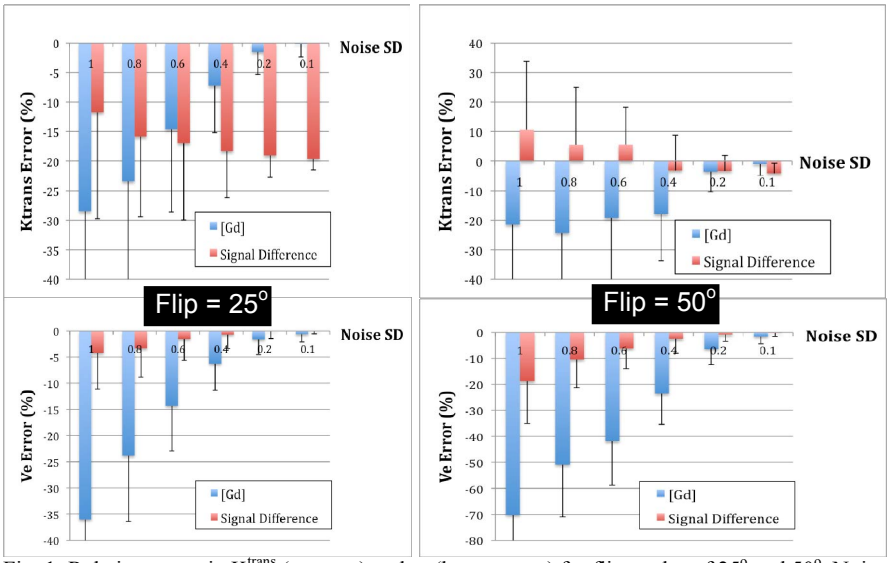

Fig. 1: Relative errors in K^{trans} (top row) and v_e (bottom row) for flip angles of 25° and 50°. Noise standard deviation is relative to the baseline AIF signal at 25° flip. So the SNR (=baseline AIF signal/noise SD) here ranges from 1 (left-most) to 10 (right-most). Shown are the average values after 100 iterations and SD error bars.

Conclusion: Our results show that under limited SNR, the signal difference method can yield more accurate perfusion measurements than the standard method of computing gadolinium concentration.

Acknowledgments: American Cancer Society RSG-08-118-01-CCE; NIH P41-RR02305; NIH R01-CA125226.

References: (1) Ashton et al. Proc. ISMRM 2007, p. 2813. (2) Walker-Samuel et al. Phys. Med. Biol. 2007, 52:589-601. (3) Ashton et al. JMRI 2008, 28:791-796. (4) Parker et al. Magn. Reson. Med. 2006, 56:993-1000. (5) Henderson et al. Magn. Reson Imaging 1998, 16:1057-1073.