Differences between DCE-MR and DCE-CT in prostate cancer and their implications on the choice of a tracer kinetics model

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Introduction

Dynamic contrast enhanced (DCE) MRI can discriminate tumor tissue in the prostate from healthy surrounding tissues [1]. The adiabatic approximation of the tissue homogeneity (AATH) model [2] and the more frequently used Tofts (modified Kety) model [3] have been used for quantification of contrast-enhanced perfusion measurements of the prostate on CT and MRI. The Tofts model has the parameter K^{trans} which has several physiological interpretations, depending on the balance between capillary permeability and blood flow [3].The AATH model on the other hand, is a distributed parameter model and is therefore able to separately determine the physiologic parameters flow (F) and extraction fraction (E). The existence of an arterial first pass peak (FPP), which is an essential part of the AATH model, is still under discussion. The question is, whether the above described advantages of the AATH model with respect to the Tofts model can be exploited on DCE-MRI quantification of the prostate. Therefore we performed a quantitative comparison of the AATH model and Tofts model in 10 patients on both DCE-CT and DCE-MRI by fitting mean contrast enhancement curves of high-flow regions suspected of tumor tissue.

Methods

All patients had biopsy proven prostate cancer and were scheduled for radiation treatment. Written informed consent was received from all patients. Approval for the study was obtained by the local ethics committee. Each patient underwent DCE multi-slice CT and DCE-MRI T1-weighed exams within one week. The DCE-CT exam was performed on a 64 slice CT scanner (120 kV, 200 mAs) and involved subsequently 24, 10 and 6 acquisitions with a time interval of 2.4, 10 and 20 s respectively. The 64 slices were reconstructed to 16 slices (2.7x2.7x2.5mm^3). After administration of the iodine contrast agent (60 ml, 6 ml/s), image acquisition started.
The DCE-MRI protocol consisted of a 3D spoiled gradient echo sequence (TR/TE=4.0/2.1 ms, flip angle 16°) using a 3T MR scanner with a two-element circular surface coil as receive coil. A total of 120 acquisitions were acquired every 2.4 s. A single acquisition consisted of 10 axial slices that were reconstructed to 1.0x1.0x5.0 mm^3 resolution. After three acquisitions, 8 ml gadolinium DTPA (0.5 M, 0.8 ml/s) was administered followed by a saline flush. Signal intensity variations were converted to changes in contrast agent concentration by using estimates of the pre-contrast T1 relaxation time, which was mapped before administration using the variable multi flip angle method (flip angles: 4.5°, 8°, 12°, 16°).
The arterial input function (AIF) was determined on DCE-CT and DCE-MRI by monitoring the signal enhancement in a central region of the left iliac externa. The AATH model was fitted to the DCE-CT data. Regions suspected of tumor tissue were defined as contiguous regions with flow higher than 20 ml / 100g min^{-1} and a minimal volume of 1 cc to reduce noise. With the aid of a rigid mutual information-based matching procedure corresponding ROIs were defined in the MR datasets. Both perfusion models were fitted to the mean enhancement curves of both regions on CT and MR. The goodness of fit is measured by the square root of the sum of squares (Q). A two-sample t-test was performed on all perfusion parameters and Q (5% significance level).

Figure 1

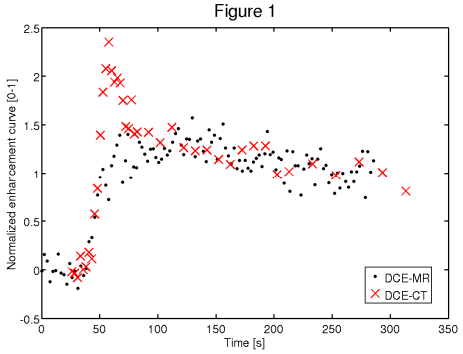


Figure 2

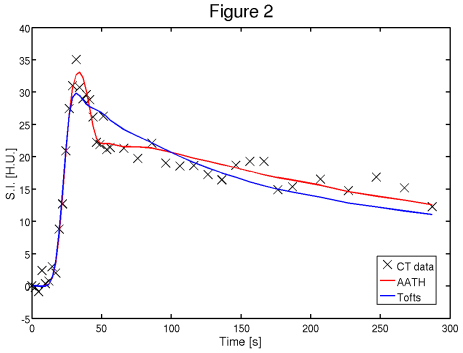
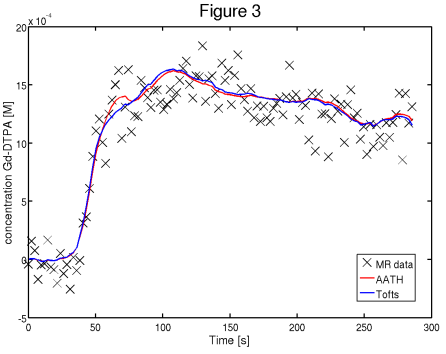


Figure 3



DCE-CT (fig. 2) and DCE-MRI (fig. 3) enhancement curves of a high-flow region (0.09cc) in one patient. Figure 1 shows the normalized enhancement curves.

Results and discussion

Eleven regions were found on DCE-CT, with an average volume of 8.5 cc (range 1.3-20.7 cc). Both models performed equal on the DCE-MRI enhancement curves, but the AATH model was able to fit the DCE-CT data significantly better than the Tofts model ($p = 0.044$, see Table 1). In 5 of 11 regions a clear FPP is visible on DCE-CT, but in none of the DCE-MRI regions, probably due to water exchange effects [4]. When looking at smaller volumes, these FPPs may show up even more distinct, as shown in figures 1-3 (volume: 0.09cc). On DCE-MRI the fit of the AATH model shows no significant improvement compared to the Tofts model, despite using two more parameters.
For the entire group, the parameters of the Tofts model are equal for DCE-MRI an the DCE-CT. However, the Tofts model fails to describe the first-pass peak on DCE-CT (fig. 2). For the AATH model, the differences between DCE-MRI and DCE-CT are remarkably small, except for the transit time. This reflects the difficulties of using the AATH model in the absence of a FPP, since transit time is very closely related to this FPP.

Conclusions

Arterial first pass peaks do exist in prostate tissue, but are not found on DCE-MRI using our protocol. So, the AATH model itself, assuming the presence of a FPP, is a valid model for perfusion analysis of the prostate. On DCE-MRI the Tofts model describes the data as well as the AATH model, whereas the AATH model describes the DCE-CT data better. For the Tofts model the parameters are equal for DCE-MRI and DCE-CT, although the model fails to describe the FPP on DCE-CT. The AATH model also yields similar results for DCE-MRI and DCE-CT, but some caution is required as the transit time shows significant differences.

TABLE 1		DCE-MRI	DCE-CT
Tofts	Q	**1.41e-4** [9.3e-5 / 2.7e-4]	**1.34** [0.59 / 2.27]
	K^{trans} (min^{-1})	**0.33** [0.13 / 0.89]	**0.31** [0.25 / 0.44]
	v_{ep}	**0.53** [0.25 / 1.00]	**0.53** [0.30 / 0.88]
AATH	Q	**1.38e-4** [9.2e-5 / 2.7e-4]	**0.92** [0.47 / 1.52]
	Flow (ml/100g min^{-1})	**31.9** [10.0 / 95.3]	**31.9** [20.5 / 46.9]
	T_c (s) *	**38.5** [10.0 / 96.0]	**24.1** [6.0 / 38.6]
	E	**0.64** [0.20 / 0.93]	**0.66** [0.43 / 0.81]
	v_{extra}	**0.49** [0.16 / 1.00]	**0.41** [0.24 / 0.75]

Mean parameter values of 11 regions (bold), range: [min / max], * = t-test, $p < 0.05$

References

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