# Extraction of an Accurate Arterial Input Function for Quantitative Perfusion and Permeability Mapping of the Prostate using an Inversion-prepared Dual-contrast sequence

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## Introduction

In dynamic contrast-enhanced MR imaging (DCE-MRI) a time series of MR images is acquired during or after injection of a low-molecular non-specific contrast medium (CM). In the first-pass bolus phase with high arterial CM concentration, a $T_2^*$-weighted pulse sequence is most sensitive to measure tissue perfusion, whereas extravasation is best depicted using a $T_1$-weighted pulse sequence. Most mathematical models for a quantitative determination of physiological and functional tissue parameters[1] require knowledge of the arterial input function (AIF), defined as the time course of CM concentration of blood entering the tissue. In the prostate, determination of the AIF in non-ECG triggered acquisitions is difficult due to strong signal intensity modulations caused by pulsatile blood flow. To get as much as possible information from DCE-MRI of the prostate, we have implemented a sequence using interleaved T1/T2* weighted acquisition, employing parallel imaging (PI) to maintain high temporal resolution. The (quantitative) AIF was calculated combining the time courses of the $T_1$- and $T_2^*$-weighted image series. Flow-induced magnitude modulations were corrected using the corresponding phase information. The method allows for simultaneous perfusion and permeability measurement in the abdomen with accurate AIF determination for application of pharmacokinetic models.

## Methods

*Acquisition:* All experiments were done on a 1.5-T whole-body scanner (Magnetom Sonata, Siemens Medical Systems, Erlangen, Germany). For DCE-MRI a transversal 5 mm slice through the prostate was chosen. A bolus of 24 ml gadopentetate dimeglumine (Magnevist, Schering AG, Berlin, Germany) was infused intravenously at a flow rate of 6 ml/s followed by 20 ml saline flush. Two FLASH images ($\alpha = 30°$, matrix size 128×90, FoV 228×228 mm²) were successively acquired every 1.65 sec: first a $T_1$- weighted image using global inversion preparation (TI = 280 ms, $TE_1 = 2.1$ ms) followed by a $T_2^*$-weighted image with $TE_2 = 27$ ms. PI was implemented using the GRAPPA technique[2] (24 reference lines, acceleration factor 2). For each echo time two dynamic series consisting of 513 magnitude and 513 phase images were reconstructed.

*Analysis:* Since the spatial sensitivity distribution of any PI coil combination is inhomogeneous, the signal intensity was retrospectively homogenized using proton-density images acquired with the body coil and the local coils respectively. The dependency of MR signal intensity on CM concentration was determined in a separate measurement in a phantom consisting of samples with different CM concentrations. To determine the AIF, the magnitude-time-courses were measured in a ROI of purely vascular voxels of the external iliac artery. Flow induced periodic signal fluctuations in the $T_1$-weighted echo series were eliminated using the respective phase images as follows: the signal variation is mainly caused by inflow enhancement. Flow velocity in the artery is correlated to the measured phase. Thus, the signal increase due to inflow can be corrected using the phase variation information. The correlation between phase and signal was determined in the equilibrium phase with slowly changing arterial CM concentration. After this correction, the AIF was fitted in both image series using a gamma variate function describing the bolus passage and a biexponential decrease to account for CM clearance. Ideally, both fits should result in the same parameters. However, at peak CM concentration the signal measured in the inversion recovery series drops due to the non-monotonic signal-concentration-dependency. Thus, the bolus length and peak were determined from fitting of the $T_2^*$-weighted echo and used as fixed input parameters for fitting of the clearance parameters in the time course of $T_1$-weighted echo.

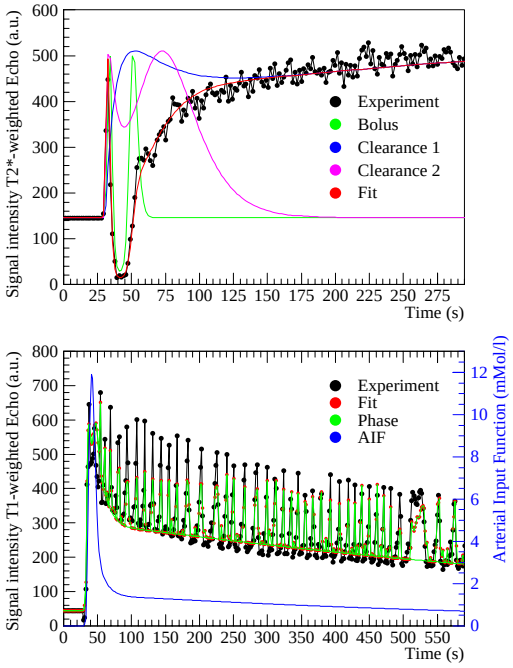


**Figure 1:** Fit of the signal-time course of the $T_2^*$-weighted (top), $T_1$-weighted (bottom) echo, and the calculated AIF (bottom).

## Results and Discussion

Figure 1 depicts the measured signals and the extracted AIF. The signal in the T2* series at peak bolus concentrations is nearly zero, suggesting that the echo time was still too long.

The phase of a voxel is mainly affected by CM concentration and flow, but not by $T_1$ relaxation enhancement. Previous studies[3] have shown that at a given CM concentration the dependency of phase on echo time is linear and that this phase velocity is linearly dependent on CM concentration. Thus, a long-term linear correction of the phase-time course after the bolus phase (t>100 sec in figure 1) accounts for the CM concentration dependency. After this phase correction, the short-term phase change is dominated by a change in flow. Linear regression between the amplitude of an arterial voxel and the corrected phase yielded a correlation coefficient of R=0.86 (P = 0.0001) and could therefore be used for eliminating the flow-induced amplitude fluctuations. Surprisingly, the correlation between the amplitude of an arterial voxel and the long-term-corrected phase was significantly better than the correlation with the ECG- or pulse signal. In DCE-MRI using $T_2^*$-weighted sequences, the degree of signal intensity loss is not only dependent on vascular CM concentration but also on micro-vessel size and density. This hampers exact quantification e.g. of the vascular volume. In $T_1$-weighted DCE-MRI signal enhancement is dependent only on the CM concentration, but the method fails during bolus passage. The implemented sequence combining both, $T_1$- and $T_2^*$-weighted DCE-MRI, enables precise quantification of the AIF at all times. It has the potential to allow for accurate pharmacokinetic modeling of signals acquired during dynamic imaging of the prostate and can differentiate if CM extravasation is either flow-limited or permeability-limited[1].

## References

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