

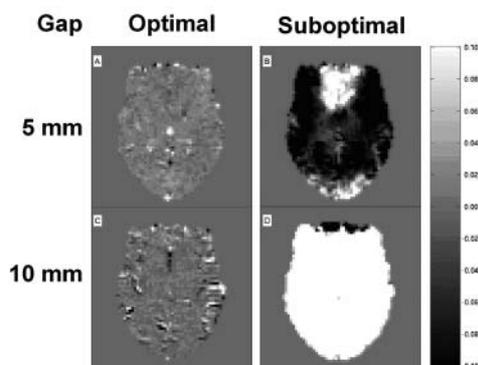
multiple presaturation pulses, without the adverse effects, by appropriate selection of crusher gradient area.

**Methods:** PICORE and FAIR measurements were carried out on a Siemens Trio 3T MR scanner with imaging parameters: TI=1000ms, FOV=196mm, 64x64 matrix, acq 100-120. The gap between inversion and imaging slices (tag gap) was 5 or 10mm. A reference scan with complete static tissue subtraction was acquired with a gap of 13mm. Presaturation pulses placed before and after the inversion pulse were used, and optimal as well as suboptimal crusher sizes were tested.

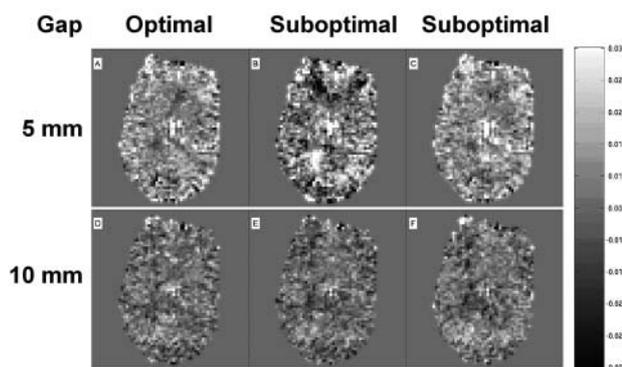
The Bloch equation was solved numerically for a range of crusher gradient areas after both the presaturation and inversion pulses. Based on the results as well as phantom measurements a set of optimal as well as suboptimal crusher values were selected for the FAIR and PICORE experiments.

**Results:**

### FAIR



### PICORE



**Figure 1.** Magnetization Difference Images

The images shown are magnetization difference images for a tag gap of 5 or 10 mm minus the difference from the reference scan divided by  $M_0$ .

With FAIR, the offset is small for optimal crusher sizes while it is very large for suboptimal crusher values (figure 1). The large offset seen in the first slice was also present in the rest of the 10 slices. In the PICORE experiment a difference between optimal and one of the suboptimal crusher values is seen if using a tag gap of 5 mm. Offsets are smaller and equal in the case of a 10 mm gap.

**Conclusion:** The choice of crusher values in FAIR can severely affect perfusion measurements. We suggest that the offset is caused by stimulated echoes. In the case of PICORE, the effects are small at a tag gap of 10 mm, but they can not be ignored for smaller tag gaps.

This study has shown that optimization of crushers can be essential for perfusion measurements if using crusher before and after the inversion pulse. Other experiments including multiple crushers before and after the inversion pulse have shown results similar to the ones in the PICORE experiment.

**References:**

[1] Sidaros, K., et al. [2003] Proceedings Eleventh ISMRM, p.2214

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### DCE-MRI liver pharmacokinetic parameters quantification by a dual-input model with non-linear sampling

**D. Monleon**<sup>1</sup>, D. Moratal-Perez<sup>2</sup>, L. Marti-Bonmati<sup>3</sup>, J. Carbonell<sup>2</sup>, B. Celda<sup>1</sup>; <sup>1</sup>Department of Physical Chemistry, University of Valencia, Burjassot, SPAIN, <sup>2</sup>BET Research Group, Universitat Politecnica de Valencia, Valencia, SPAIN, <sup>3</sup>Department of Radiology, Dr. Peset University Hospital, University of Valencia, Valencia, SPAIN.

**Purpose:** The blood supply to the liver is derived jointly from the hepatic arteries and the portal venous system. The large number of pharmacokinetic parameters and the fast variations in contrast concentration in the first seconds after a paramagnetic contrast media injection reduce the efficiency of linear sampling. Non-linear sampling seems to be required to optimise the measured points in dynamic contrast enhanced (DCE) MR imaging.

**Subjects and Methods:**

**Subjects:**

A complete protocol of DCE-MR with parametric pharmacokinetic analysis was applied to 20 subjects. A contrast agent (0.5 mM) was randomly injected at either 4 ml/s or 5 ml/s and an amount of 0.2 or 0.3 ml/Kg.

**Pulse sequence program:**

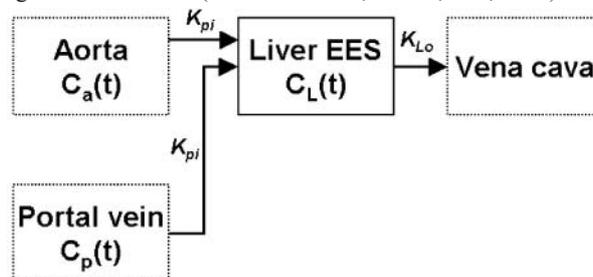
The MR series (24 slices covering the whole-liver, 13 dynamic acquisitions) were acquired in 210 seconds, each dynamic lasting 4 sec. The dynamics were non linearly distributed along the theoretical liver perfusion curve. The pulse sequence used included two blocks of 4 consecutive acquisitions during a breath hold, between 20 and 60 seconds after injection. The remaining 5 acquisitions were distributed along the whole duration of the experiment from the moment previous to bolus injection (precontrast series) to the final point at steady-state.

**Co-registration of images:**

4D (XYZ + time) co-registration of MRI images was achieved by iterative application of built-in routines of the SPM2 (United Kingdom) package.

**Pharmacokinetic model:**

A one-compartment two-input model was used to describe the liver perfusion (Fig 1). Parameters for optimisation include  $k_{pi}$ ,  $k_{ai}$ , and  $k_{lo}$ . Curves-representation, area-integrations, Levenburg-Marquard least-squares and parametric imaging were obtained using MATLAB 6.5.1. (The Mathworks, Natick, MA, USA).



**Figure 1.** Dual-input one-compartment model for liver perfusion.

**Results:** Parametric images from 20 subjects were obtained by curve fitting, area integration and parameter optimization of non-linear sampling DCE-MRI data. Non-linear time sampling of liver perfusion provided more precise contrast concentration curves than regular linear sampling. Errors in 4D co-registration greatly contributed to the uncertainty in the final results.

**Discussion/Conclusion:** The non-linear sampling used in this work allowed more precise definition of the contrast agent concentration curves and better parametric determination in liver perfusion studies. Good 4D (XYZ + time) co-registration is required to apply pharmacokinetic models to DCE-MRI data.

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### Impact of SSFP - banding artefacts on quantitative perfusion imaging

H. Köstler, C. Ritter, D. Baunach, D. Hahn, J. Sandstede; *Institut für Röntgendiagnostik, Universität Würzburg, Würzburg, GERMANY.*

**Purpose:** Perfusion imaging using a saturation recovery steady state free precession (SR-SSFP) sequence suffers from dark banding artefacts during the peak contrast agent concentration. Aim of the study was to investigate the dependence of the artefact on the location and whether the artefact may be misinterpreted as a perfusion deficit in quantitative perfusion imaging.

**Methods:** 21 first-pass perfusion examinations (3 slices each, SR-SSFP sequence, 5 x 3 ml Gd-DTPA, 5 x 9 ml Gd-DTPA, 5 x 12 ml Gd-DTPA, 6 x 1 ml / 8 ml Gd-DTPA pre bolus-technique [1]) were performed in healthy volunteers at rest. 8 sectors per slice were evaluated. Quantitative evaluation used contamination correction [2] and deconvolution with the arterial input function. Artefact levels (al) were classified visually as not present: 0, visible: 1 and substantial: 2.

**Results:** The artefact was located at the interface of blood in the ventricle and the myocardium. Number and extent of the artefacts increased with higher contrast agent dose (table 1). The artefacts were less pronounced in the mid-ventricular slice (table 2) and strongest in the septum (table 3, septum: SE, posterior wall: PW, lateral wall: LW, frontal wall: FW). The measured perfusion increased with higher artefact level (artefact level 0: 0.65 ml/min/g, artefact level 1: 0.72 ml/min/g, artefact level 2: 0.77 ml/min/g).

**Discussion:** Dark banding artefacts are quite common in SR-SSFP perfusion imaging, especially in the septum of basal and apical slices. This leads to difficulties in the determination of the onset of the signal increase and to a steeper slope. However, this results in increased perfusion values and thus the SSFP - banding artefact can not be misinterpreted as a perfusion deficit in quantitative perfusion imaging.

#### Literature:

[1] Köstler H, Ritter C, Lipp M, Beer M, Hahn D, Sandstede J [2003] Abstract: MAGMA 11 supplement 1: Pre bolus technique for MR quantification of myocardial perfusion.

[2] Köstler H, Ritter C, Reiss-Zimmermann M, Beer M, Hahn D, Sandstede J [2004] Magn. Reson. Med. 51: 848 - 852: Correction for Partial Volume Errors in MR Heart Perfusion Imaging.

table 1: artefact level dependence on the contrast agent dose

	3 ml	9 ml	12 ml
al 0	60	68	47
al 1	43	29	43
al 2	17	23	30

table 2: artefact level dependence on the slice position

	base	mid	apex
al 0	62	105	57
al 1	66	46	63
al 2	40	17	48

table 3: artefact level dependence on the wall segment

	SE	PW	LW	FW
al 0	31	51	78	64
al 1	41	49	34	51
al 2	54	26	14	11

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### Detection of blood flow reserve in peripheral arterial disease with Muscle-BOLD

D. von Elverfeldt<sup>1</sup>, E. Buschmann<sup>2</sup>, S. Schirmer<sup>2</sup>, M. Buechert<sup>1</sup>, J. Hennig<sup>1</sup>; <sup>1</sup>Radiology, Medical Physics, University Hospital Freiburg, Freiburg, GERMANY, <sup>2</sup>Cardiology and Angiology, University Hospital Freiburg, Freiburg, GERMANY.

**Introduction:** Recently it has been shown that the well known BOLD-contrast is present in muscle tissue also and can be observed e.g. directly after exercise [1]. In addition, patients with peripheral vascular occlusive diseases show a delayed BOLD signal [2]. In this study the blood flow recovery of patients suffering from peripheral vascular obstructive disease (PAD) was monitored via muscle-BOLD during arteriogenesis treatment.

**Materials and Methods:** Five patients with symptomatic PAD Rutherford I Category 3-3 were included in the study. The patients received Granulocyte-Macrophage-Colony-Stimulating-Factor (GM-CSF) subcutaneously in 2day intervals over 14days. The MR-measurements were performed on days 0, 14 and 90.

The patients exercised in the scanner by repeatedly pressing a foot pedal with nearly isometric muscle contraction until the onset of ischemic pain occurred, while the control group exercised as long as possible. The subsequent time resolved BOLD-measurement employed a fat suppressed EPI sequence with TE=60ms, TR=818ms and 185 scans.

Three proximal lower limb muscles were manually segmented for data analysis. The time courses were evaluated as the mean signal intensity of the muscle volumes and fitted with a Boltzman step function modified with an exponential decay. The position and slope of the inflection point calculated from the fit function was taken as a surrogate parameter of the BOLD response (see Fig.2).

**Results:** The patient group showed a significantly delayed inflection point and, therefore, a delayed BOLD response for the gastrocnemius muscles (Fig.3). The slope of the BOLD response from the musculus soleus in the patient group showed an increase during the treatment comparing day zero and day 90 as seen in Fig.4. The change in slope of the BOLD-response for the two gastrocnemius muscles is not statistically significant, yet both muscles show an increase similar to the soleus muscle.

**Discussion:** The potential diagnostic value of this method was demonstrated by the significant differences in the muscle-BOLD response between patients and normal volunteers. However, the variations of the derived parameters, e.g. the involuntary use of different muscle groups, still restrict a reliable diagnosis. The measurement of a flow response to reactive hyperemia instead of physical exercise might improve the accuracy significantly. This hypothesis will be subject to further investigations. However, mus-