The three different phase patterns with their corresponding target-states in transverse plane (Mx, My). The first one (1) divides the read direction in four areas, with respectively a magnetization phase equal to 0, pi/2, pi and 3pi/2. The second one (2) defines 4 areas in which magnetization phases are equal to 0, pi, 0 and pi. The last target state (3) defines 3 areas: the phase of the first one increases linearly from 0 to pi, the phase of the second one is equal to pi, and the phase of the last one decreases from pi to 0. The bottom row presents the three B1 pulses (magnitude and phase) resulting from the application of the Pontryagin Maximum Principle.

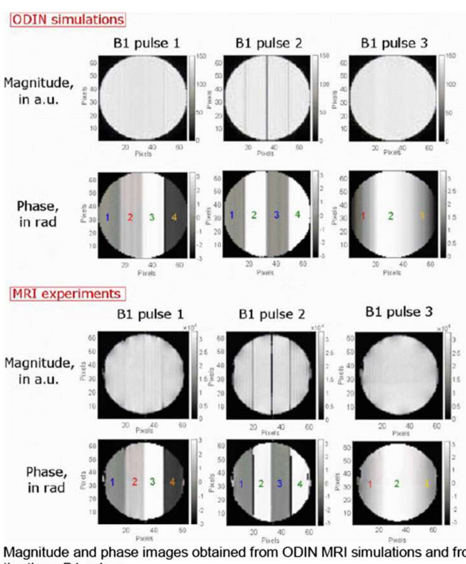
Results: Experiments were performed on a homogeneous agar phantom (concentration 1.5% in weight), whose relaxation times were T1 = 1500 ms and T2 = 130 ms. A numerical phantom having similar properties was generated for simulation.
Simulations were performed with the ODIN MRI simulator[4]. MRI experiments were run on a Bruker 4.7T MRI system. Parameters of the sequences used in simulations and experiments are presented in Fig. 2. A spin-echo sequence was used where the excitation pulse was replaced by the OC-designed pulse.

Parameters of the sequences used in simulations and experiments.

Parameters	ODIN simulation	MRI experiment
Sequence	Spin-echo	Spin-echo
Matrix	64 x 64	64 x 64
FOV	2.0 x 2.0 cm	3.25 x 3.25 cm
Echo time	6.6 ms	9.8 ms
Repetition time	5 s	5 s
G_{OC}	11.7 mT/m	8 mT/m

Parameters of the sequences used in simulations and experiments.

Figure 3 shows the results of the ODIN simulations and MRI experiments with both magnitude and phase images.



Magnitude and phase images obtained from ODIN MRI simulations and from MRI experiments, with the three B1 pulses.

Discussion/Conclusion: In both simulations and experiments, the target-states are reached: phase images are consistent with Fig. 1.
In cases 1 and 2, phase images from simulations and experiments present sharp transitions between the different areas: this demonstrates that the B1 pulses accurately control the phase, with good frequency selectivity.

This work illustrates the feasibility to control magnetization phase with OC-designed pulses and is a first step to apply OC to phase-based MRI techniques, such as diffusion or MRE. One advantage of this method could be to relaxe the constraints on having oscillating gradients in these both techniques.

References:
1. L. S. Pontryagin, Mathematical theory of optimal processes, CRC Press, 1987.
2. E. Van Reeth, H. Ratiney, M. Tesch, D. Grenier, O. Beuf, S.J.Glaser, D Sugny, 'Optimal control design of preparation pulses for contrast optimization in MRI', Journal of Magnetic Resonance, vol. 279, pp. 39–50, 2017.
3. N. Khaneja, T. Reiss, C. Kehlet, T. Schulte-Herbrüggen, and S.J.Glaser, 'Optimal control of coupled spin dynamics: design of NMR pulse sequences by gradient ascent algorithms', Journal of Magnetic Resonance, vol. 172, no.2, pp. 296–305, 2005.
4. T. H. Jochimsen, A. Schäfer, R. Bammer, and M. E. Moseley, "Efficient simulation of magnetic resonance imaging with Bloch–Torrey equations using intra-voxel magnetization gradients," Journal of Magnetic Resonance, vol. 180, no. 1, pp. 29–38, 2006.

Acknowledgements: We acknowledge support from the ANR-DFG research program Explosys (Grant No. ANR-14-CE35-0013-01; GL203/9-1). This work was performed within the framework of the LABEX PRIMES (ANR-11-LABX-0063/ANR-11-IDEX-0007).

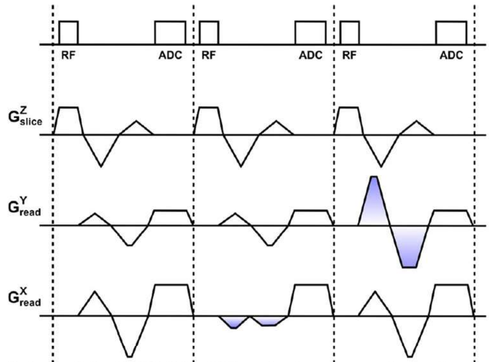
305

2D In-plane Flow MRI in Real Time

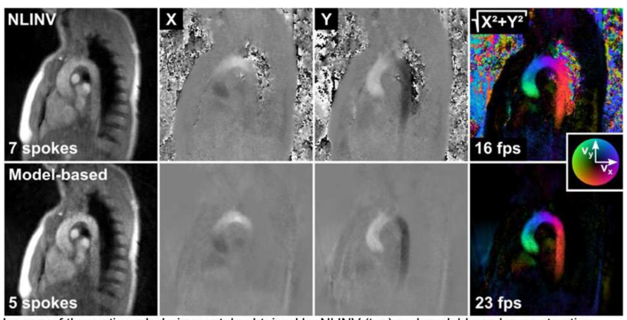
J.M. Kollmeier, A.A. Joseph, D. Voit, Z. Tan, K.-D. Merboldt, J. Frahm
Biomedizinsche NMR Forschungs GmbH, Max-Planck-Institut für biophysikalische Chemie, Göttingen/GERMANY

Purpose/Introduction: Phase-contrast (PC) flow MRI has successfully been adapted to real-time MRI based on undersampled radial FLASH [1,2] with image reconstruction by nonlinear inversion (NLINV) [3] and a model-based approach [4]. So far, however, flow-encoding for these real-time techniques has been limited to through-plane flow, neglecting in-plane flow components. As a first step towards multi-dimensional flow encoding, this study realized 2D in-plane flow imaging in real time.

Subjects and Methods: Experiments were performed on a 3 T MRI system (Prisma, Siemens Healthcare, Erlangen, Germany) using an 18-channel thorax coil in combination with suitable elements of the spine coil. A highly undersampled radial FLASH PC sequence was developed using in-plane flow sensitivity, flow compensation on all axes (Fig. 1) and a 30% asymmetric echo [2] to minimize TE. PC velocity maps for each flow direction individually were obtained by both NLINV with subsequent complex subtraction of differently flow-encoded images and a model-based reconstruction. Other parameters: VENC = 200 cm s^{-1}, TR/TE = 2.94 ms/2.09 ms, flip angle 10°, in-plane resolution 1.5 mm, slice thickness 6 mm, FOV 320 mm, 3 × 7 and 3 x 5 radial spokes. The achievable measuring times per dataset were 62 ms and 44 ms, respectively.



Undersampled radial 2D PC flow MRI merging flow-compensating and flow-encoding gradients (blue).



Images of the aortic arch during systole obtained by NLINV (top) and model-based reconstruction (bottom). (From left to right:) Magnitude image, PC velocity maps in A>>P direction and in H>>F direction, and resulting speed image with color-coded flow direction.

Results:
The image series (movies) for both reconstruction methods show quantitative velocity maps at high spatiotemporal resolution that demonstrate the successful realization of real-time 2D flow MRI. The PC maps of the aorta reveal the expected velocity pattern at high SNR with minimal artifacts. Figure 2 shows single systolic frames selected from a continuous movie covering multiple heart beats. The benefit of the model-based reconstruction is apparent not only by the suppression of phase noise but furthermore by an improved spatiotemporal accuracy. The speed image with color-coded flow direction allows for a combined representation of both flow components permitting online visualization.

Discussion/Conclusion: This study demonstrates that real-time 2D flow MRI can be realized by using highly undersampled radial FLASH with NLINV or model-based reconstruction. At this stage, 2D-flow movies achieved a temporal resolution of 44 ms without the need for prospective or retrospective gating. The present work focused on in-plane flow, while foreseeable extensions to 3D-flow MRI are expected to result in measuring times of less than 60 ms.

References:
1. Joseph AA et al., NMR Biomed 25:917–924, 2012
2. Untenberger M et al., Magn Reson Med 75:1901–1908, 2015
3. Uecker M et al., Magn Reson Med 63:1456–1462, 2010
4. Tan Z et al., Magn Reson Med 77:1082–1093, 2016

306

Simultaneous Multi-Contrast (SMC) Imaging for Synchronous DWI and T2*-Weighting

N.-J. Breutigam[1], R. Frost[2], K. Eickel[1], D. Porter[1]
[1]MR Physics, Fraunhofer MEVIS, Bremen/GERMANY, [2]Radiology, Athinoula A. Martinos Center for Biomedical Imaging, Boston/ UNITED STATES OF AMERICA

Purpose/Introduction: The multiple image contrasts required in clinical diagnostic MRI are usually acquired as sequential scans. Acquisition times for these scans can be reduced by using simultaneous multislice (SMS) imaging[1]. However, in many clinical protocols, the potential scan-time reduction is not fully exploited because the repetition time (TR) cannot be shortened without compromising image quality. Simultaneous multi-contrast (SMC) imaging[2] uses methodology from SMS to reduce examination times without a reduction in TR. This paper reports an updated application of SMC to the combined acquisition of diffusion-weighted (DW) and T2*-weighted (T2*W) images; this combination of contrasts is of particular interest in acute stroke[3].

Subjects and Methods: *Pulse Sequence:*
Measurements were performed using a DW readout-segmented, echo-planar (rs-EPI) sequence[4−5] with blipped CAIPIRINHA[6−7], which was modified to acquire signal with different contrasts from two slice positions at the same time. The first slice was used to generate DW contrast, whilst the second slice provided T2*W contrast (see Fig. 1).

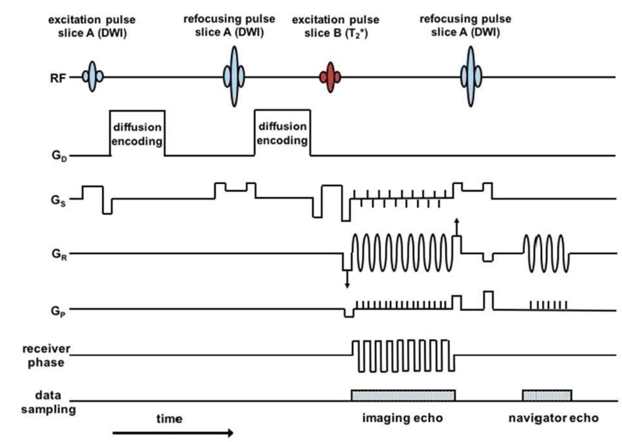


Figure 1: Pulse diagram of the rs-EPI sequence for SMC imaging. Data are acquired from two slice positions at the same time with one slice (A) generating DW while the other slice (B) provides T2*-weighted contrast. Firstly, slice A is excited and DW preparation is applied. Then slice B is excited before both signals are sampled simultaneously using rs-EPI with a variable amplitude encoding gradient in the readout (G_R) direction (labelled with arrows) and a blipped phase-encoding gradient (G_P). A blipped-CAIPIRINHA[6-7] gradient scheme along the slice-select (G_S) direction is used in conjunction with receiver phase modulation to shift the T2*-weighted image by half a field of view (FOV) in the phase-encoding direction relative to the DW image. Finally, an RF refocusing pulse is applied to slice A only to generate a 2D navigator signal to phase correct the DW imaging data.

rs-EPI Pulse Diagram for SMC