Self-gated Arterial Spin Labeling Perfusion Mapping of Myocardium Using Magnetic Resonance Imaging

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Abstract

Myocardial perfusion MR provides useful information for diagnosis of several cardiac diseases and allows for treatment monitoring. Many methods have been proposed based on DCE-MRI or ASL. With the current slow drift from contrast agent administration in clinical setup, it is necessary to develop reliable methods that would replace DCE-MRI. ASL has always been a hope for a clean non-invasive method of perfusion measurement, but due to its issues it is still not widely accepted as a reliable method especially outside of the brain.

We propose a method for perfusion measurement based on FAIR-ASL which internally does not require ECG measurement, uses advanced acquisition tweaks and compressed sensing subspace image reconstruction. We test the method in a preclinical setup, where high B0 fields and higher heart rates are the main issues.

The acquired flow maps were comparable with state of the art methods, while giving a possibility of acquisition shortening or increased spatial resolution. We were able to obtain perfusion maps of the whole cardiac cycle with 315 um spatial resolution and 1 mm slice thickness from a 15-minutes acquisition. The acquisition time could possibly be reduced further, by additional optimization of the pulse sequence.

1. Introduction

Myocardial perfusion magnetic resonance (MR) provides a useful biomarker for diagnosis of cardiac diseases and treatment monitoring. Multiple methods have been proposed, mainly based on first-pass Dynamic Contrast-Enhanced Magnetic Resonance Imaging (DCE-MRI) [1] or Flow-Alternating Inversion Recovery Arterial Spin Labeling (FAIR ASL) [2–4]. These methods suffer from low reliability due to the low signal-to-noise ratio (SNR) of the MR data, need for Electrocardiographic (ECG) triggering, breath-hold acquisition or no agreement on unified data processing. This is even more challenging in small-animal MRI, with lower SNR and much higher cardiac and respiratory rates than in humans. Additional problems arise from generally higher B0 field (7T+) and higher gradient-system strengths used for small animal imaging, where the magnetohydrodynamic effect and noise from gradient coils basically prevent ECG measurement during scanner operation.

Therefore we propose a method based on our previous work [5], which attempts to overcome most of these uncertainties by incorporating retrospectively self-gated acquisition with advanced compressed sensing imaging acquisition and reconstruction.

2. Methods

The method described in this paper is based on the self-gated T1 mapping sequence with the FAIR module derived from [5], a 2D inversion-recovery method with spoiled-gradient-echo (SPGR) readout and radial k-space sampling. Special care must be taken when selecting an appropriate inversion pulse as the FAIR experiment relies on perfect and global inversions. In standard setups this is hard to achieve as the length of the homogeneous-B1 region in the excitation RF coil is usually shorter than rat's (or human) body length. The pulses provided by the scanner manufacturer suffer from poor adiabacity and low bandwidth. Therefore on-the-fly computation of Shinnar–Le Roux (SLR) derived adiabatic pulses [6] was implemented. These pulses allow, thanks to lower peak values than for example standard hyperbolic secant RF pulses, for shorter pulse durations, thereby providing a wider bandwidth at a similar level of adiabaticity. On-the-fly correction for gradient imperfections was employed in order to compensate for non-ideal slice selection [7].

The dataset is split into two experiments, one with selective and one with global inversion pulse preparation. Both datasets are then binned to select appropriate respiration and cardiac phases (diastole and systole). During the acquisition it is ensured to start the inversion pulse at the

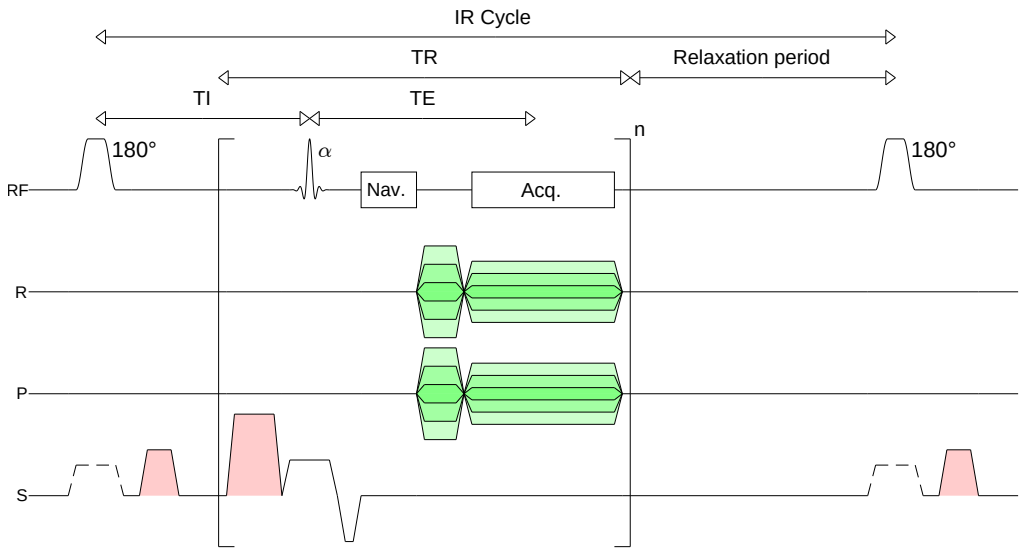

Figure 1. 2D golden-angle Spoiled Gradient Echo FAIR pulse-sequence timing diagram. The navigator is acquired within the "Nav." block.

same phase of the respiration cycle, based on prospective gating using a standard respiration pillow. In the image-reconstruction step, the radial k-space trajectory is corrected using the measured gradient transfer function [8] and the images were reconstructed using the BART reconstruction toolbox [9].

For image reconstruction a subspace based image reconstruction method [2, 10] was selected. To construct the subspace a model derived from [11] was used assuming non ideal inversion, imperfect B1+ homogeneity, two compartments with possibly distinct T1 values and arbitrary contribution to the overall signal by the two compartments [4, 12]. In total 1 000 000 curves were generated from which a subspace consisting of 6 basis functions was generated, retaining 99 % of information (chosen by the magnitude of the singular values). Spatial total variation (TV) regularization was used during the image reconstruction in order to simultaneously denoise the data. The coil sensitivities were estimated using the NLINV tool based on Non-linear inversion [13]. The curves were later fitted with the TOMROP model [11] to obtain the T1 values for the selective and non-selective cases. The perfusion maps were computed according to [4], modeling also the effect of the inversion pulse on blood in cardiac ventricles in the selective-inversion case.

3. Animal experiment

All measuring procedures were performed under the EU Directive no. 2010/63/EU and approved by the Animal Care Committee of Czech Academy of Sciences, Czech Republic, and Czech Governmental Animal Care Committee, in compliance with Czech Animal Protection Act No. 246/1992. The imaging was performed on a 9.4T Bruker Biospec 94/30 (Ettlingen, Germany) preclinical MR scanner, equipped with the BGA12S-HP gradient system. A volume resonator with an inner diameter of 86 mm was used as an excitation coil alongside a 4-channel array surface coil for reception. A series of anatomical images were acquired to locate the short-axis view of the myocardium. A single slice was positioned intersecting with papillary muscles in the left ventricle where the perfusion was measured. The acquisition parameters were: Matrix size: $192 \times 192 \times 1$, FOV: $40 \times 40 \times 1$ mm, 1700 excitations per inversion cycle, repetition time between excitations was set to 6 ms with flip angle of 4° and relaxation delay at the end of readout train was set to 5 seconds. The acquisition length was set to 15 minutes during which both selective and non-selective experiments were performed in sequential manner. That means several global inversion cycles followed by the same number of selective inversion cycles.

4. Results

The measured MBF values were estimated in systole to be 490 ± 108 ml/100g/min. The spread in the values was quite noticeable but usual as with other methods . There was no visual difference between the observed MBF in the anterior and posterior left-ventricular wall perfusion and T1 values (Fig. 2).

As a byproduct of the proposed method also global T1 maps were generated with T1 in myocardium of 1550 ms, which can serve as an additional biomarker.

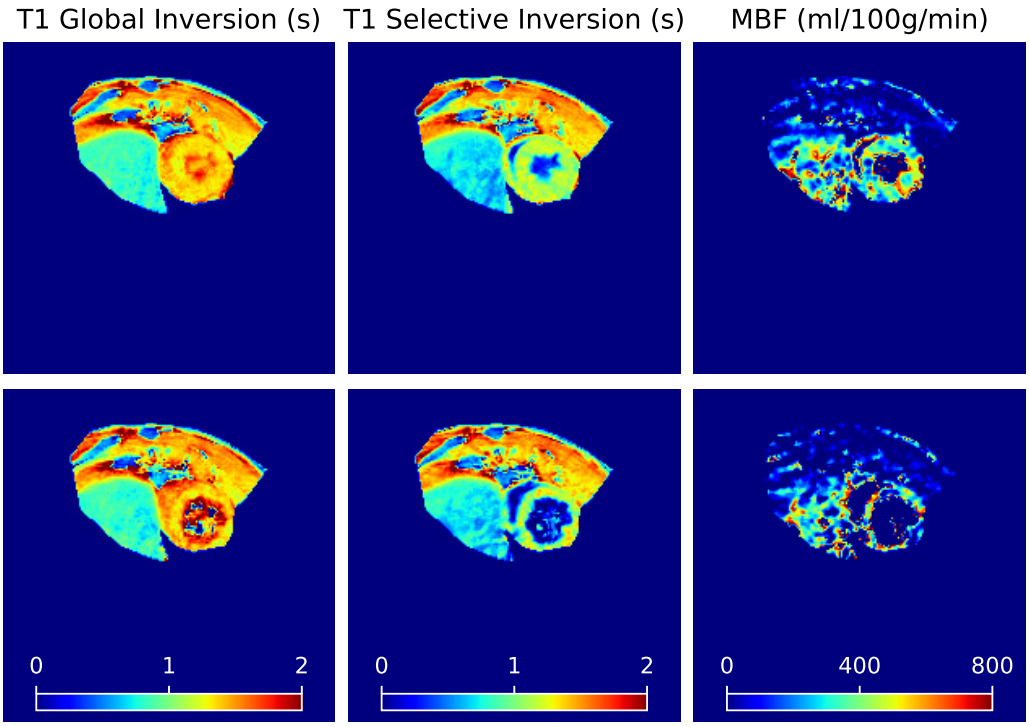

Figure 2. Resulting selective and non-selective T1 maps and Myocardial blood flow (MBF) map from a healthy rat. Top row systole, bottom row diastole.

5. Discussion

The acquired perfusion values at rest (under anesthesia) are comparable to values reported in literature [3, 14](500 and 600 ml/100g/min), though it must be stated that absolute values depend heavily on the applied anesthetics its amount and the evaluated region.

To evaluate detection of changes in pathological states, animals need to be measured under various conditions affecting MBF to show the ability of our method to detect perfusion changes, which is the natural extension of this conference paper and will be pursued in the near future. As with ASL in general SNR is the main issue, the possibility to tune the regularization weights within the image reconstruction pipeline can help in denoising the image at the cost of details. Further tuning of the regularization weight might be necessary to lower the spread in the MBF values.

Since there was no visual inhomogeneity in the T1 values and perfusion values it indicates that omitting ECG triggering during the experiment did not introduce any further errors. This is possibly due to the fact that the slice-selective inversion slab is thicker than the imaging slice. So even if the heart moves, the imaging slice still stays within the inversion slab.

The possibility of extracting T1 global maps from the ASL measurement can shorten standard clinical protocols for measurement of extracellular fractional volume (ECV) in fibrotic studies and provide additional information (MBF) for little extension in scan protocol time.

Previously [5] we have shown that this T1 mapping sequence is quite resilient concerning B1 inhomogeneity thanks to low flip angle. In this experiment we've chosen a slightly higher flip angle in order to get more T1 weighing and higher SNR. This might cause lesser resilience to B1 inhomogeneities, therefore it should be investigated more in the follow-up work. On the other hand the measured global T1 values in myocardium did not differ, when compared to the ones mentioned in [5] and others. This could indicate that by giving a longer relaxation period in this experiment, we did not give up much of B1 inhomogeneity resilience after all.

It is possible that the developed method could be used also in measuring perfusion in other areas, where movement is at play. Most often ASL is performed in brain, where little to no movement is present. On the other hand our method theoretically opens a way to examine for example tumor perfusion in the proximity of heart or in the abdominal cavity, where mainly breathing artifacts are of concern.

6. Conclusion

We showed a method for measuring myocardial blood flow without the use of a contrast agent, which is based on the self-gated Look-Locker FAIR ASL principle. The resulting MBF maps corresponded to the ones reported in literature. Our method showed the possibility of measuring the blood flow with higher spatial resolution than for example [2] for little increase in measurement time and without the need for ECG triggering increasing animal welfare and experimentator comfort. The method could be possibly used in other areas of interest, where motion artifacts are a common issue.

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