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Adiabatic spin-lock preparations enable robust in vivo cardiac $T_{1\rho}$-mapping at 3T

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Abstract—Magnetic Resonance Imaging (MRI) is the clinical gold standard for the assessment of myocardial viability but requires injection of exogenous gadolinium-based contrast agents. Recently, $T_{1\rho}$-mapping has been proposed as a fully non-invasive alternative for imaging myocardial fibrosis without the need for contrast agent injection. However, its applicability at high fields ($> 1.5$ T) is hindered by susceptibility to MRI system imperfections, such as inhomogeneities in the $B_0$ and $B_1^+$ fields.

In this work we propose a single breath-hold ECG-triggered single-shot bSSFP sequence to enable $T_{1\rho}$-mapping in vivo at 3T. Adiabatic $T_{1\rho}$ preparations are evaluated to reduce $B_0$ and $B_1^+$ sensitivity in comparison with conventional spin-lock (SL) modules. Numerical Bloch simulations were performed to identify optimal parameters for the adiabatic pulses. Experiments yield $T_{1\rho}$ values in the myocardium equal to $148.13 \pm 54.08$ ms for the best adiabatic preparation and $16.01 \pm 20.75$ ms for the reference non-adiabatic SL, with $26.91\%$ against $89.74\%$ relative difference in $T_{1\rho}$ values across two imaging conditions. Both phantom and in vivo measurements show increased myocardium/blood contrast and improved resilience against system imperfections compared to non-adiabatic $T_{1\rho}$ preparations, enabling the use at 3T.

Clinical relevance—Adiabatically-prepared $T_{1\rho}$-mapping sequences form a promising candidate for non-contrast evaluation of ischemic and non-ischemic cardiomyopathies at 3T.

I. INTRODUCTION

Cardiac Magnetic Resonance Imaging (MRI) is the main clinical tool for tissue characterization of the myocardium. Specifically, Late Gadolinium Enhancement (LGE) is the clinical gold standard for the assessment of myocardial viability. In LGE, gadolinium-based contrast agents are administered and retained in scarred tissue about 10 minutes after injection. This leads to shortened $T_1$ times compared to healthy myocardium and induces a contrast with the surrounding healthy tissue. However, the need for exogenous contrast agents limits its repeated use. Furthermore, its use is contraindicated in patients with renal dysfunction due to the risk of nephrogenic systemic fibrosis (NSF) and recently gadolinium retention in the brain has been observed, raising the level of caution.

The emergence of myocardial parameter mapping offers a promising pathway to contrast-free assessment of myocardial viability. However, initial attempts based on $T_1$ and $T_2$-mapping did not provide sufficient contrast for the robust assessment of scar and fibrosis in clinical MRI [1]. $T_{1\rho}$-mapping has recently been proposed as a promising endogenous contrast alternative for imaging myocardial fibrosis [1] - [6]. $T_{1\rho}$ is the time constant that characterizes the longitudinal magnetization relaxation in the rotating frame of reference. $T_{1\rho}$ relaxation is induced by the continuous application of an RF pulse, the so-called spin-lock (SL) pulse [7]. As a result, the magnetization is locked along the RF field, mitigating the loss of transverse magnetization and, thus, suppressing the low frequency contribution to the relaxation that would lower the contrast between normal and infarcted myocardium. SL preparations with different durations are used to generate varying $T_{1\rho}$ contrast in MRI scans. The data acquired with different $T_{1\rho}$ weightings are then fit to an exponential decay function to obtain voxel-wise estimation of $T_{1\rho}$ values.

While promising results were obtained at low field strength [6], the translation of $T_{1\rho}$-mapping to higher field strengths is hindered by the susceptibility of the SL preparation to $B_0$ and $B_1^+$ field inhomogeneities. Adiabatic SL modules, consisting of amplitude and frequency modulated RF pulses, have the potential to overcome these limitations and allow the use of $T_{1\rho}$ as an endogenous contrast agent in clinical practice at high fields [8], [9]. In this work, we sought to enable robust in vivo $T_{1\rho}$-mapping at 3T in a single breath-hold using adiabatic SL preparations. Bloch simulations were performed to optimize the parameter choice for adiabatic preparations. The proposed sequence was then tested on agar-based phantoms and on a healthy volunteer to study the performance of adiabatic and conventional SL preparations in presence of system imperfections.

II. METHODS

A. Sequence design

$T_{1\rho}$-mapping was performed on a 3T scanner (Ingenia, Philips, Best, The Netherlands). Four $T_{1\rho}$-prepared images are acquired followed by a saturation-prepared image to approximate infinite SL duration and capture the effects of the imaging readout (Fig. 1). $T_{1\rho}$-prepared images were acquired with different total SL duration ($\tau_{SL} = \{0, 60, 120, 180\}$ ms) to induce varying $T_{1\rho}$ contrast and interleaved with 3 s delay to allow for $T_1$ recovery. A single-shot balanced Steady-State Free Precession (bSSFP) sequence was used for the acquisition, with the following imaging parameters: Flip Angle = 70°, phantom: resolution = $1 \times 1 \times 8$ mm$^3$, FOV =
Fig. 1. (A) T1ρ-mapping sequence acquiring four T1ρ-prepared images, interleaved with 3 s delay to allow T1 recovery, and a saturation-prepared image in a single breath-hold. (B) Reference continuous-wave Spin-lock (SL) preparation. (C-D) Adiabatic SL preparations consisting of a train of 2 or 4 HS pulses, with equal total duration.

250 × 180 mm², TE/TR = 1.08/2.73 ms, in vivo: resolution = 1.8 × 1.8 × 8 mm³, FOV = 260 × 190 mm², TE/TR = 0.95/2.25 ms. Cardiac motion compensation was achieved by ECG triggering the acquisition to the mid-diastolic phase. The total scan time was 13 ms, performed during a single breath-hold to mitigate respiratory motion.

B. T1ρ preparations

Two adiabatic T1ρ preparations were compared to a non-adiabatic SL module (RefSL) [6]. The conventional SL module consists of a 90° tip-down pulse followed by an RF pulse with constant amplitude and frequency (duration = τSL). Finally, a 90° tip-up pulse is applied to restore the longitudinal magnetization. The RefSL block presents additional phase alternation to partially compensate for B0 inhomogeneities. Adiabatic T1ρ preparations, on the other hand, are obtained by concatenating 2 (HS-2) or 4 (HS-4) phase-cycled hyperbolic-secant adiabatic full-passage pulses (Fig. 1b-d) [10]. These pulses are characterized by variable amplitude and frequency, where:

\[ B_1(t) = B_1^{\text{max}} \cdot \text{sech} \left( \beta \left( \frac{2t}{\tau_{HS}} - 1 \right) \right) \]

\[ \Delta \omega_1(t) = f_{\text{max}} \cdot \tanh \left( \beta \left( \frac{2t}{\tau_{HS}} - 1 \right) \right) \]

The duration of each pulse was determined by the total SL block duration of 60 ms (τHS = 30 ms for HS-2, τHS = 15 ms for HS-4). The peak RF amplitude was set to the maximum (B1+max = 13.5 μT). Bloch simulations of the magnetization evolution for different configurations of adiabatic SL preparations were performed to optimize the remaining parameters β and fmax. Assuming that an ideal adiabatic preparation yields a final magnetization in alignment with the z-axis, the preparation efficiency was computed through the ratio Mz/Mz(0) of the final and initial longitudinal magnetization, where Mz(0) = 1. Average Mz was computed for every combination of \( \beta = 1, 1.25, 1.5, \ldots, 10 \) and \( f_{\text{max}} = 0, 50, 100, \ldots, 5000 \) Hz over a design window of ±150 Hz off-resonances and ±25% \( B_1^+ \) variations [11].

C. Phantom and in vivo experiments

The sequence was first tested in a NiCl2-doped agar-filled vials phantom, submerged in a water bath. Each vial contained a different agar concentration to achieve two different shimming conditions to test the resilience of the different preparations to system imperfections. Voxel-wise T1ρ maps were generated for phantom and in vivo scans, using a three-parameter exponential decay model. In vivo T1ρ maps were segmented to extract the left ventricular myocardium and blood pool.

III. RESULTS

The simulation results for HS-2 and HS-4 preparations averaged over the design window present periodic patterns across β and fmax values. For both pulses the best preparation efficiency was obtained for low to intermediate frequency sweep amplitudes and showed an inversely proportional relationship with the parameter β (Fig. 2a-b). Overall, HS-2 shows a higher preparation efficiency in the optimal region than HS-4. Optimal values of \{β, fmax\} were chosen as \{3, 500 Hz\} for HS-2 and \{3.5, 450 Hz\} for HS-4, resulting in average residual magnetization \( M_z \) of 0.98 and 0.96, respectively. Examples of preparation efficiency over a range
of ±200 Hz $B_0$ and 0% − 200% $B_1^+$ inhomogeneities are shown for selected parameter combinations (Fig. 2c-f).

Fig. 3a shows phantom magnitude images for the longest SL duration and the final $T\rho_1$-maps for RefSL ($f = 500$ Hz), HS-2 ($f_{\text{max}} = 250, 500$ Hz), and HS-4 ($f_{\text{max}} = 450$ Hz). HS-2 (500 Hz) preparations yield the best results, with fewer artifacts than both HS-4 and RefSL, higher $T\rho_1$ contrast than non-adiabatic preparations, and less susceptibility to the $B_0$ shim. The mean ± standard deviation $T\rho_1$ values over the three artifacts-free vials shown in Fig. 4 are vial 1, vial 2, vial 3 = 384.95 ± 40.04, 373.42 ± 39.24, 108.19 ± 9.87 ms for HS-2, 299.31 ± 56.30, 293.12 ± 91.21, 69.66 ± 8.37 ms for HS-4 and 104.96 ± 87.25, 98.12 ± 81.40, 54.64 ± 48.71 ms for RefSL. Non-adiabatic and adiabatic $T\rho_1$ dispersion results in Fig. 4a show consistent trends, with $T\rho_1$ values increasing with $f$ and $f_{\text{max}}$, respectively. Overall, adiabatically-prepared sequences yield longer $T\rho_1$ times compared with non-adiabatic pulses.

Fig. 5 shows the short-axis $T\rho_1$-weighted baseline images and the corresponding overlaid $T\rho_1$-maps for the myocardium and left-ventricular blood pool of a healthy volunteer. In vivo average $T\rho_1$ values (± standard deviation) in the myocardium were: 16.01 ± 20.75, 148.13 ± 54.08, and 54.72 ± 41.04 ms for RefSL, HS-2, and HS-4 preparations, respectively (Fig. 5). Adiabatically-prepared sequences show significantly lower noise and higher contrast between the myocardium and the blood pool than RefSL. Moreover, HS-4 preparations show more $T\rho_1$ inhomogeneities across the myocardium and significant artifacts for shimming 2. Specifically, the differences in measured myocardial $T\rho_1$ values between the two shimming conditions are 89.74%, 26.91%, 58.03% for RefSL, HS-2, and HS-4, respectively (expressed in % over the average $T\rho_1$ obtained through shimming 1).
Those areas show artificially high preparation efficiency values, which result from very fast adiabatic sweeps that do not induce any rotation in the magnetization. Accordingly, in phantom experiments, HS-2 preparations show reduced artifacts in the water bath compared to HS-4 preparations. In vivo HS-2 maps are artifact-free for both shimming conditions. HS-4 maps, on the other hand, present residual off-resonance artifacts as well as more inhomogeneous $T_1\rho$ values across the myocardium. These results indicate that longer HS pulses, with slower frequency sweeps, achieve better performance and thus HS-2 preparations are to be preferred over MLEV-cycled HS-4 modules.

V. CONCLUSIONS

Our results suggest that adiabatic $T_1\rho$-preparations allow for robust in vivo quantification of myocardial spin-lock (SL) relaxation times at high field strengths. This paves the way for potential contrast-free imaging of myocardial fibrosis at 3T.

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