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Inversion Recovery at 7 Tesla in the human myocardium: measurement of T_1 , inversion efficiency and B_1^+

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Abstract

At clinical MRI field strengths (1.5 and 3T), quantitative maps of the longitudinal relaxation time T_1 of the myocardium reveal diseased tissue without requiring contrast agents. Cardiac T_1 maps can be measured by Look-Locker inversion recovery sequences such as ShMOLLI at 1.5 and 3T. Cardiovascular MRI at a field strength of 7T has recently become feasible, but doubts have remained as to whether magnetization inversion is possible in the heart because of subject heating and technical limitations. This work extends the repertoire of 7T cardiovascular MRI by implementing an adiabatic inversion pulse optimized for use in the heart at 7T. A “ShMOLLI+IE” adaptation of the ShMOLLI pulse sequence has been introduced together with new post-processing that accounts for the possibility of incomplete magnetization inversion. These methods were validated in phantoms and then used in a study of 6 healthy volunteers to determine the degree of magnetization inversion and the T_1 of normal myocardium at 7T within a 22 heartbeat breathhold. Using a scanner with 16 kW radiofrequency outputs, inversion efficiencies ranging from -0.79 to -0.83 (intra-segment means; perfect 180° would give -1) were attainable across the myocardium. The myocardial T_1 was $1925 \pm 48\text{ms}$ (mean $\pm$ SD).

Keywords

7 Tesla; Cardiac longitudinal relaxation time T_1 ; ShMOLLI; Adiabatic pulses; Inversion recovery; Saturation correction

Introduction

Since its invention, MRI has moved to ever-stronger magnetic fields to exploit the concomitant increases in signal-to-noise ratio and the opportunity to obtain images with improved contrast. Despite formidable challenges from the underlying physics, cardiac MRI

has recently been demonstrated at 7 Tesla (7T) (1). The FLASH (fast low-angle shot) (2) and bSSFP (balanced steady-state free precession) (3) images obtained are of sufficient quality to assess cardiac volume and function (2-6); and right coronary angiography is feasible (7). Human cardiac imaging at 7T is an exciting area that is still in its infancy (8,9).

The spin-lattice relaxation time T_1 plays a central role in determining the acquisition parameters leading to optimal contrast in MR images. In vivo, it is only feasible to determine T_1 empirically despite a mature theory (10). The T_1 of human myocardium at 7T is worth measuring because it is important for imaging, impossible to calculate and no experimental value has been reported in the literature. In cardiac MRI, T_1 is often measured using an inversion recovery (IR) pulse sequence employing the Look-Locker scheme (11,12) with ECG triggering. In a single short breath hold, modern pulse sequences such as MOLLI (13) and ShMOLLI (14) measure maps of the T_1 for every pixel in an image. The resulting quantitative parametric maps of T_1 differentiate normal and diseased tissue (14,15). Magnetization inversion, which is central to measuring T_1 , is also a key component in many of the cardiac pulse sequences that are used clinically at established field strengths. Examples include sequences for selective suppression of fat, blood and healthy myocardium; angiography; myocardial tagging; and viability imaging (late-Gd enhancement) (16).

The primary aim of this work was to measure the T_1 of normal human myocardium at 7T for the first time. To achieve this, an adiabatic inversion pulse was optimized to perform magnetization inversion in the human heart at 7T. This was incorporated into a variant of the ShMOLLI (14) pulse sequence and post-processing algorithms, which were adapted to determine the extent to which inversion was successful and to mitigate the effects should inversion be incomplete.

Theory

An important goal of this work was to quantify the effectiveness of inversion pulses in the myocardium to inform future work at 7T. An inversion pulse is characterized by the ratio of the magnetization immediately after the inversion pulse $M_z(TI=0)$ to the equilibrium magnetization M_0 . This inversion efficiency $IE = M_z(TI=0)/M_0$ varies between -1 for perfect inversion, 0 for saturation and +1 for a 0° pulse.

T_1 is typically determined from Look-Locker IR data by a 3-parameter fit of signals in each pixel at time TI after inversion originally described by Deichmann and Haase (17):

$$s(TI) = A - B \exp(-TI/T_1^*) \quad [1]$$

with fitted parameters A , B and T_1^* . Providing the readout flip angle is small, this empirical time constant T_1^* is then corrected for partial saturation to give

$$T_1^{DH} \approx T_1^* / [A/M_0] \approx T_1^* / [A/(B-A)] = T_1^* (B/A - 1). \quad [2]$$

ShMOLLI is a Look-Locker inversion recovery (IR) pulse sequence that produces T_1 maps in the heart at 1.5T and 3T from 7 single-shot images acquired during a 9 heartbeat breath hold (14). ShMOLLI reconstruction uses magnitude images, employing heuristics for each pixel to infer the sign of M_z and decide whether to include the 6th and/or 7th images in the fitting. Data are then fitted to Eq. [1] and corrected for saturation using Eq. [2] to yield T_1 maps.

However, the derivation of Eq. [2] assumes perfect inversion $IE = -1$ so that the initial magnetization $M_z(TI=0) = -M_0$ and hence $M_0^{DH} \approx B - A$. Repeating that derivation without this assumption gives

$$\tilde{T}_1 \approx T_1 * \frac{A - B}{A \times IE} \quad \text{and} \quad \tilde{M}_0 \approx \frac{A - B}{IE}. \quad [3]$$

Comparing this result with Eq. [2], we see that $T_1^{DH} = \tilde{T}_1 \times -IE$ so that any imperfections in the inversion pulse will have a direct bearing on measurements made according to Eq. [2]. At 7T, technical challenges in delivering high B_1^+ power mean that one cannot presume that inversion will be perfect nor even constant across the heart. It is, therefore, essential to measure the extent of magnetization inversion in order to obtain a reliable value for the myocardial T_1 at 7T. We introduce two complementary approaches to determine this.

The first method is to measure the equilibrium magnetization directly, by acquiring an additional image at least $5 \times T_1$ after inversion. With a heart rate of 70bpm, image 8 in Fig. 1 occurs 13 heartbeats after the third inversion and fulfils this criterion when $T_1 \approx 2.2$ s. Hence

$$T_1^{Im8} \approx T_1 * [A/M_0] \approx T_1 * s_{Im8}/A. \quad [4]$$

A slower but more accurate approach fits a 4-parameter model to each pixel s_i in the 12 complex-valued images. The model assumes that $|M_z(t)| \propto |s_i|$ for each pixel in each image and infers the sign of $M_z(t)$ from the pixel phase as follows. Since image #8 samples the fully relaxed spins $M_z(t_8) = 0$ and hence $M_z < 0$ in other images whenever the phase difference $|\arg(s_i) - \arg(s_8)| > \pi/2$. For computational efficiency, inversion is taken to occur instantaneously at the middle of each inversion pulse and readout instantaneously during the central k-space line. Spoiling of transverse magnetization is assumed to be complete so M_x and M_y may be neglected. In summary, the 4 parameters are: relaxation time T_1^{4P} , equilibrium signal $A^{4P} \propto M_0$, inversion efficiency IE^{4P} and the fraction of M_z preserved after each readout F^{4P} . Figure 1 shows an example of M_z and s_i according to this model.

Methods

Experimental methods

At 7T, peak B_1^+ is limited and B_0 - and B_1 -shimming leave appreciable residual field inhomogeneity. Pilot experiments, in phantoms and in vivo at 7T (18), motivated several changes to ShMOLLI to sidestep these obstacles. Figure 1 shows the resulting “ShMOLLI +IE” pulse sequence, which enabled us to measure both T_1 and the efficiency of

magnetization inversion. The modifications were as follows: (a) A choice of adiabatic inversion pulses was provided in order to optimize magnetization inversion in vivo; (b) FLASH readout was used instead of bSSFP to reduce off-resonance effects and minimize radiofrequency (RF) heating; and (c) an 8 heartbeat delay and 5 images without inversion preparation were added to measure the equilibrium magnetization M_0 and readout-induced saturation. In total, 12 complex-valued images were acquired in a single, longer, end-expiration breath hold. The heart-rate dependent timings were recorded. Each complex-valued image was converted by the scanner into a pair of corresponding magnitude and phase images, which were stored as DICOM files for post-processing. All experiments used a Magnetom whole-body 7T MRI scanner (Siemens Healthcare, Germany) with custom RF hardware capable of transmitting up to 8kW (in single channel mode) or 16×1 kW (in B_1 -shimming mode). Validation used a phantom comprising 7 tubes containing water, agarose, Gd(DTPA)^{2-} , 20mM choline chloride, 20mM sodium acetate and a dash of sodium azide. The tubes were immersed in water in a 1000mL, 104mm Ø, plastic jar (Nalgene). Concentrations of agarose and Gd(DTPA)^{2-} were chosen to yield four different T_1 s and also four different T_2 s (in tubes with $T_1 \approx 840$ ms). The reference T_1 and T_2 were measured by multi-point IR-SE and SE experiments with $\text{TR}=10$ s. An 8kW rated head coil with 24 receivers (Nova Medical Inc., MA, USA) was used for validation measurements to allow a high B_1^+ to be produced in some validation experiments to ensure that inversion pulses could be observed in the adiabatic limit and to facilitate reliable spin-echo reference measurements. Without moving the phantom, ShMOLLI+IE scans were then performed with a systematic variation of the sequence parameters matching the in vivo experiments detailed below.

Ten volunteers were recruited in accordance with IRB approval from the University of Minnesota. Technical problems with adapting prior techniques to the newly-installed scanner required 4 scans to be abandoned. The remaining six male volunteers (21-29y, 59-82kg, 1.60-1.88m) completed matching protocols using a 16-element TEM transceiver array (19,20) driven in B_1 -shimming mode (21) by 16×1 kW RF amplifiers (Dressler, Germany). The specific absorption rate (SAR) was monitored (22) to ensure it was within the limits set forth by the U.S. Food and Drug Administration. The first level limit for local SAR would permit up to 40W kg^{-1} to be delivered during a 10s interval. Maximum local SAR was $0.285 (\text{kg}^{-1}) \times \text{input power (W)}$ (23). Hence, a maximum of 97ms at full power (16kW from the amplifiers) would have been acceptable over a 10s interval. ECG triggering was used and each volunteer was B_1 -shimmed (24,25) over a mid-short-axis slice and B_0 -shimmed over an $8 \times 8 \times 8 \text{cm}^3$ cube around the heart (WIP package 452B, Siemens Healthcare, Germany).

The ShMOLLI+IE scans of each subject used readouts in mid-diastole (~ 475 ms after R-wave), with single shot FLASH, linear k-space ordering, 6/8 partial Fourier, $2 \times$ GRAPPA, 144×100 matrix and a $340 \times 240 \times 8 \text{mm}^3$ field of view. On the basis of Bloch simulations (see SI Figs. 1-6) and pilot experiments (not shown), the following sequence parameters were varied systematically. Inversion used a 10ms hyperbolic secant (HS) pulse (26) (the same as traditional ShMOLLI), HS $_n$ pulses with $n=1, 2, 4, 6$ and 8 (27) and a B_1 -independent rotation (BIR4) pulse (28). HS $_n$ pulses with higher n are expected to achieve inversion with

a lower B_1^+ but to have more ripple when B_1^+ exceeds this adiabtic onset (see SI Fig. 6). The HS8 pulse was tested with durations $T_p=5, 10, 20$ and 30ms (the maximum duration the RF amplifier could produce), and with time-bandwidth products $R=2, 5, 8, 10, 20, 30$ and 40 . Longer pulses are expected to produce more effective inversion until T_2 relaxation becomes significant during the time of the pulse (see SI Fig. 3 and (29)). Pulses with a smaller time-bandwidth product are expected to produce more effective inversion but only in tissues with a narrow range of frequency offsets (see SI Figs. 4-5). To test for the onset of adiabaticity, the HS8 pulse was tested with peak RMS voltages of $100, 150, 200\text{V}$ and the maximum voltage ($\approx 210\text{V}$ depending on the volunteer) (30). The readout nominal flip angle was varied between 2° and 30° . Measurements with the set of parameters thought most likely to produce effective inversion (HS8, $T_p = 10\text{ms}$, $R = 10$, maximum power and 5° nominal FLASH readout) were repeated at the beginning and end of the session to gauge reproducibility. Each volunteer contributed more than 30 ShMOLLI+IE scans to the analysis.

Analysis

Offline analysis was all performed in Matlab (MathWorks, Natick, MA) using the “lsqcurvefit” routine from the Optimization Toolbox for non-linear fitting. The IR-SE reference data were processed by least-squares fitting each pixel to $S(TI)=A - B \exp(-TI/T_1)$ where $S(TI)$ is the signal in the image acquired at time TI after the inversion pulse and A, B and T_1 are parameters to be estimated. To determine T_2 , each pixel in the SE reference data was fitted to $s(T_E)=A \exp(-T_E/T_2)$ where T_E is the echo spacing.

Data acquired with ShMOLLI+IE were fitted to Eq. [1] using traditional ShMOLLI reconstruction and corrected for saturation using Eq. [4]. The raw data were also processed offline with the 4-parameter fit described above. In both cases, the coefficient of explained variance R^2 was calculated for each pixel, serving as a quality control for adherence of the data to the model.

To evaluate the resulting parametric maps, regions of interest (ROIs) were drawn manually using only the magnitude images so as not to be swayed by the measured T_1 s. In the phantom, one ROI was drawn for each phantom tube. Following convention (31), the myocardium in each mid-short-axis slice was outlined and divided into ROIs covering the 6 segments. Pixels which did not fit the model ($R^2 < 0.97$) were excluded before calculating the mean and standard deviation (SD) in each ROI. Averaging over all the data, 1% of pixels were excluded from T_1^{Im8} analysis in the septum increasing to 9% at the rear of the heart, while 33% of pixels were excluded from T_1^{4P} analysis in the septum increasing to 81% at the rear of the heart. Excluded pixels were observed to concentrate along the blood-myocardium boundary or in areas where the myocardium had been difficult to segment manually.

Results

Phantom experiments (Figs. 2-4 and SI Figs. 7-8) were performed to validate the ShMOLLI+IE pulse sequence and analysis. Particular attention was devoted to characterize potential

errors arising from incomplete inversion, incomplete recovery before inversion or saturation during readout.

To investigate incomplete inversion, the inversion pulse peak voltage was reduced progressively (Figs. 2A and 3). The observed time constant T_1^* was almost independent of inversion efficiency, but deviated significantly from the true T_1 due to readout-induced saturation. “Correction” according to Eq. [2] gave T_1^{DH} , which was close to the reference T_1 at high voltages where inversion was effective, but which seriously underestimated T_1 when inversion was incomplete. T_1^{Im8} was more accurate than T_1^{DH} at all voltages; T_1^{4P} was best of all. Both T_1^{Im8} and T_1^{4P} avoided introducing systematic errors due to incomplete inversion, although effective inversion still had the benefit of minimizing the ROI standard deviation.

To investigate the impact of magnetization recovery, data were acquired with a range of simulated heart rates (Fig. 2B). T_1^* showed noticeable changes, which were eliminated by all three post-processing methods. Again, T_1^{4P} was most accurate, having –3% error for R–R intervals 600 – 1500ms.

Finally, the readout flip angle was varied to investigate readout-induced saturation (Fig. 2C). With increasing readout flip angle, T_1^* became much shorter than the true T_1 due to saturation. T_1^{DH} and T_1^{Im8} both eliminated half the readout-induced error, while T_1^{4P} was essentially impervious to readout-induced saturation, except that smaller readout flip angles had the benefit of minimizing the ROI standard deviation. The robust performance of T_1^{4P} continued over the range of T_1 and T_2 in the phantom (Fig. 4A).

These experiments also assessed the measured inversion efficiencies. The ROI mean IE^{Im8} and IE^{4P} were compared against a Bloch simulation while the inversion pulse peak voltage was changed (Fig. 3). IE^{4P} agreed more closely with the simulations and had smaller ROI SDs. Nevertheless, IE^{Im8} was of acceptable quality when the voltage was sufficient to make $IE < 0$. In terms of magnetization recovery, IE^{4P} changed $< 0.3\%$ and IE^{Im8} changed $< 4\%$ over R–R intervals of 600–1500ms (data not shown). Finally, the efficiency of the inversion pulse at each phantom tube should depend only on the local B_0 , B_1^+ and T_2 (29) but *not* vary with readout flip angle. Yet measured inversion efficiency was found to be more sensitive than T_1 to readout flip-angle (Fig. 4), demonstrating the limits of accuracy of inversion efficiency determined by the 4-parameter fit as flip angle increases. However, when the flip angle was kept below 10° , IE^{4P} in different phantom tubes was stable and in agreement with Bloch simulations. Figure SI 7 (see Supplementary Information) further illustrates the reliability of with a 5° FLASH readout.

Together, these results demonstrate that ShMOLLI+IE using 5° FLASH readout and 4-parameter fitting allowed robust, accurate measurement of T_1 and IE in phantoms. Although less accurate, T_1^{Im8} and IE^{Im8} were also useful because they are available immediately during scanning.

Moving to human subjects, Fig. 5 illustrates results for a mid-short-axis view of the heart. The left ventricle is clearly visible. However, in the blood pool the 4-parameter fit does not

converge and T_1^{Im8} is much lower than expected from the literature (32) and is not consistent with anatomy.

A range of inversion pulse parameters was evaluated through an extensive series of measurements of inversion efficiency. Measurements were made in vivo (Fig. 6), in order to guarantee realistic field inhomogeneities, coil loading, physiological motion, etc. The HS8 pulse with $R = 10$, $T_p = 10\text{ms}$, and 210V peak voltage had the best inversion and also the lowest SD; the segment mean IE^{4P} was -0.83 to -0.79 . Readout flip angles $2-30^\circ$ had no significant effect on IE^{4P} (not shown). These trends are consistent with theory (33) and phantom data (not shown due to lack of space).

Figure 7 summarizes the myocardial T_1 measurements. Figure 7A demonstrates a progressive failure of traditional T_1^{DH} analysis when inversion becomes incomplete e.g. at low transmit power; yet T_1^{4P} and T_1^{Im8} remain consistent in spite of this. Figure 7B shows that T_1^{4P} is most resilient against readout-induced saturation errors. Figure 7C summarizes all available mid-short-axis acquisitions using HS8 pulses with 5° nominal readout flip angle (117 in total). Points marked in blue used optimal parameters, from which the segment mean T_1^{4P} had mean 1925ms ($\pm 48\text{ms}$ inter-segment SD) and the segment mean T_1^{Im8} had mean 1880ms ($\pm 63\text{ms}$ inter-segment SD). Black points show T_1^{4P} values from the range of experiments used to investigate IE^{4P} in Fig. 6. It is apparent that T_1^{4P} is robust to sub-optimal inversion just as observed above in phantoms. Equivalent analysis (not shown) gave group mean T_1^{Im8} within 5% of Fig. 7C, but having up to 70% greater SD.

Finally, we note that the measurements of inversion efficiency vs inversion pulse voltage described above may be used to estimate B_1^+ . Figure 8A shows a map of B_1^+ obtained by fitting the phantom IE^{4P} data pixel-by-pixel to an HS8 pulse Bloch simulation. The simulation depended weakly on T_2 , which for simplicity was therefore taken to be 81ms everywhere. An actual flip-angle imaging (AFI) (34) B_1^+ map was also acquired. Figure 8C compares each pixel of the ShMOLLI+IE and AFI B_1^+ maps. There is close agreement except near the edge of the phantom, where $B_1^+ < 3\text{ HzV}^{-1}$ and the AFI map can be seen to break down. An equivalent analysis was applied in vivo to the segment mean IE^{4P} values. Figure 8D shows B_1^+ in the heart in vivo, taking an approximate value for the myocardial T_2 because this has not yet been measured at 7T.

Discussion

The ShMOLLI+IE pulse sequence, shown in Fig. 1, was designed to measure robustly T_1 and inversion efficiency in the heart at 7T. It addresses the principal challenges: that peak B_1^+ is limited by the cost of power amplifiers, RF heating (SAR) limits the duty cycle and B_1 -inhomogeneity is inevitable when imaging the body at 7T (25). B_0 inhomogeneity around the heart-lung interface is also worse at 7T than at 3T (35). The optimization of adiabatic inversion pulses (shown in Fig. 6) is a trade-off between the B_1^+ power and B_0 homogeneity necessary for inversion across the heart. As predicted by Bloch simulations (see SI Figs. 1-6), the best performance at 7T was found for a 10ms HS8 pulse which, compared to the 10ms HS pulse used at 3T, required lower B_1^+ power to achieve inversion although it also required stricter B_0 homogeneity (27). Thus, we used a prototype B_0 -

shimming package to reduce the bandwidth of signals from the myocardium, making it possible to invert magnetization with the RF hardware available. Meanwhile, FLASH imaging was chosen for the readouts because it is more robust against artifacts than bSSFP and because it contributes less SAR. Indeed, the RF amplifier duty cycle and T_2 relaxation already limited the inversion pulses to ~ 30 ms, so SAR did not restrict our choice of FLASH readout or HS8 inversion parameters.

Theory, phantom and in vivo experiments (Eq. [3], Figs. 3 and 7A) all confirm the critical importance of perfect inversion for accurate T_1 measurements using the Deichmann-Haase correction T_1^{DH} . Yet it is impossible to achieve perfect inversion in vivo because of T_2 relaxation and limited peak B_1^+ . The impact of inversion efficiency when measuring T_1 has been discussed for the brain (36,37). Quantifying and correcting for imperfect inversion is also critical to the 7T cardiac experiments presented here. Moreover, theory predicts up to $\sim 5\%$ error in cardiac T_1 due to incomplete inversion at lower fields. Although 4-parameter fitting is introduced here in the context of experiments at 7T, it should also be applicable at 1.5 and 3T.

Four-parameter fitting is a compromise between the empirical model in Eq. [1] and a full Bloch simulation of the magnetization, which would require measurement of several tissue parameters simultaneously such as the tissue T_2 , B_0 , B_1^+ , B_1^- , extent of flow and the through-slice tissue distribution. Bloch simulations support these approximations for the relevant range of parameters. In particular, the relatively short myocardial T_2 and use of spoiler gradients in the sequence allows M_x and M_y to be neglected. Fitting in Matlab on a desktop PC ran in <5 min without needing optimization of the code.

It is instructive to compare our findings with data from the established ShMOLLI protocol at 1.5 and 3T. The average healthy myocardial T_1 measured by ShMOLLI+IE is 1925 ± 48 ms (inter-segment mean $\pm$ SD). This compares with average healthy myocardial T_1 s by ShMOLLI at 1.5T of 1006 ± 50 ms (mean $\pm$ SD) and 1215 ± 63 ms at 3T (14). (These figures include correction of the known 4% underestimation of T_1 by ShMOLLI. See Figs. 3A and 3B in reference (14) for details.) Our findings continue the trend of increasing T_1 at high field strength, as predicted by theory in cases where dipole-dipole coupling dominates T_1 relaxation (10). The subject-average inversion efficiency achieved by the optimized HS8 inversion pulse was between -0.83 and -0.79 here. At 1.5 and 3T, it is assumed that the inversion efficiency is -1 . However, Bloch simulations suggest that the inversion efficiency of ShMOLLI's 10ms HS pulse is closer to -0.90 to -0.95 , which might explain ShMOLLI's known 4% underestimation of T_1 .

We have presented the first pulse sequence for measuring myocardial T_1 at 7T. However, a number of practical limitations remain to be overcome before 7T cardiac T_1 mapping is ready for use in clinical research. Most obviously, ShMOLLI+IE requires a 22 heartbeat breath hold compared to 9 heartbeats for ShMOLLI. The mean imaging time per subject here was $19 - 27$ s compared to 9.0 ± 1.1 s for ShMOLLI (14), although we note that our volunteers all indicated that the long breath hold was not challenging. Furthermore, we anticipate that in future this limitation will be removed by re-ordering the scan such that the equilibrium magnetization is sampled *before* the first inversion pulse. The 8 heartbeat

recovery interval would no longer be required. The details of this optimization are intended to be presented elsewhere in due course.

Using the 16 element TEM coil and B_1 -shimming, it was only possible to deliver the high B_1^+ required for efficient inversion over a small volume of tissue covering the heart. Furthermore, it was also necessary to concentrate the B_0 optimization over a similar small volume. Together, this meant that although magnetization inversion was effective in the heart, the inverted region did not extend uniformly across the abdomen as would be the case for body-coil transmission at 1.5T. Analogous to spin-labeled perfusion methods such as FAIR (38), this means that regions with in-flowing blood will demonstrate the expected T_1 recovery during the heartbeat in which inversion occurs, followed by a more rapid magnetization recovery during the ensuing heartbeats. For fitting according to Eq. [4], this gives the false appearance of a short T_1 which is visible in Fig. 5A as a region of apparently low T_1 in the left-ventricular blood pool. During 4-parameter fitting, this mismatch between the apparent T_1 in images at short TI compared to those from subsequent heartbeats leads to a failure to converge, which is visible in the blood pool in Figs. 4C-D.

The experiments reported here were performed using custom-built RF hardware, which produced B_1^+ of at least 260Hz in the heart at full power. As a guide to the prospects for magnetization inversion in the heart using other 7T systems, SI Fig. 9 takes data for the power-dependence of inversion efficiency from Fig. 6d and scales the x-axis using the B_1^+ data from Fig. 8d. It is apparent that magnetization inversion demands significant B_1^+ , which we achieved here using a purpose-built RF coil and B_1^+ -shimming. Looking to the future of cardiac MR at 7T, our experience here suggests that parametric mapping is feasible, that inversion pulses are not yet as effective as at 3T even with the best hardware, that careful optimization of inversion pulses is critical, that increased RF transmit power and improved B_0 shimming play complementary roles and that sequences relying on inversion of spins over an extended volume such double-inversion-recovery preparation (“dark blood”) will be difficult to implement.

Conclusions

Inversion pulse performance plays an important role in the determination of T_1 . Correction for imperfect inversion was necessary for reliable measurement of T_1 in the heart at 7T. The combination of ShMOLLI+IE and 4-parameter fitting permitted measurements of T_1 and IE that were validated against standard reference methods in phantoms and were consistent over a wide range of sequence parameters in vivo. It was indeed possible to produce inversion in the heart at 7T; the best performance was with an HS8 pulse of duration $T_p=10\text{ms}$ and time-bandwidth product $R=10$. Using this pulse, inversion was reasonably effective and relatively uniform across the myocardium (intra-segment mean IE = $-0.79 - -0.83$). The myocardial T_1 at 7T was found to be $1925 \pm 48\text{ms}$ (inter-segment mean $\pm$ SD) compared to $1215 \pm 63\text{ms}$ at 3T and to $1006 \pm 50\text{ms}$ at 1.5T (from (14) including the 4% correction previously described).

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.

Schematic diagram of the “ShMOLLI+IE” pulse sequence introduced here. **a:** The sequence is triggered from ECG R waves. **b:** The first 9 heartbeats closely resemble traditional ShMOLLI, except that the adiabatic inversion pulse (“180°”) has been changed and the readouts (grey boxes) use FLASH instead of bSSFP. Addition of a delay and 5 further readouts permits evaluation of equilibrium magnetization and readout-induced saturation. **c:** Illustration of the longitudinal magnetization $M_z(t)$ and signal intensity s_i for one pixel during the ShMOLLI+IE pulse sequence when $T_1^{4P} = 1.5 \times \text{R-R-interval}$, $A^{4P} = 1.0$, $IE^{4P} = -0.9$ and $F^{4P} = 0.7$. (These parameters are defined in the Theory section).

Fig. 2.

Phantom validation of T_1 measured with ShMOLLI+IE and the four post-processing algorithms described in the Theory section. Data were acquired using a head RF coil with high B_1^+ efficiency, 5° FLASH readout, 100V HS8 10ms R=10 inversion pulse and $T_R=1$ s except as indicated on the horizontal axes. Phantom tube T_1 (mean $\pm$ SD) is compared with the reference $T_1=1835$ ms (black line) and $T_2=81$ ms. Connecting lines are drawn only to guide the eye. T_1^{4P} is most resistant to incomplete magnetization inversion (a), incomplete magnetization recovery (b) and readout-induced saturation (c).

Fig. 3.

Phantom validation of inversion efficiency (ROI mean $\pm$ SD) measured with ShMOLLI+IE in the phantom tube with reference $T_1 = 1835\text{ms}$ and $T_2 = 81\text{ms}$. Experiments used a head RF coil for high B_1^+ efficiency, 5° FLASH readout, HS8 10ms $R=10$ inversion pulse and T_R was 1s. A Bloch simulation (black line) was fitted to IE^{4P} with a single adjustable parameter $B_1^+ = 6.8 \text{ Hz V}^{-1}$. The simulation predicts a best-case IE = -0.95 with a 70V pulse (black $\times$).

Fig. 4.

Extended phantom validation of four-parameter fitted T_1^{4P} and IE^{4P} . Data were acquired as described in Figs. 2–3, but the analysis was extended to cover all 7 phantom tubes to study the impact of the reference T_1 s and T_2 s shown in the legend. **a:** T_1^{4P} (mean $\pm$ SD) for each tube in comparison with reference T_1 s (lines). **b:** IE^{4P} (mean $\pm$ SD) for each tube. Points marked "Sim." show the minimum M_z from Bloch simulations of the HS8 pulse using the reference relaxation values in the legend (analogous to the $\times$ in Fig. 3). For clarity, connecting lines are drawn to guide the eye and small horizontal offsets were applied for readout flip angles $\geq 10^\circ$.

Fig. 5.

Illustrative examples of T_1 and inversion efficiency IE maps from a single ShMOLLI+IE acquisition in the heart at 7T. **a-b:** Analysis with Eq. [4]. **c-d:** Analysis with 4-parameter fitting. In all panels, pixels where the data do not fit the model well ($R^2 < 0.97$) are plotted semi-transparent to avoid distracting the reader. Dashed lines show the endo- and epicardial contours with radial lines indicating the right ventricular insertion point, *i.e.* between the anterior (#7) and anteroseptal (#8) segments (31). The other segment boundaries fall at 60° intervals, radiating from the centroid of the epicardial contour.

Fig. 6.

Optimization of inversion pulse parameters in vivo. The parameters used were: 10ms HS8 R=10 inversion pulse with 5° FLASH readout, except with the variations specified on each subplot. Points show inter-segment inversion efficiency IE^{4P} for 36 myocardial segments (mean $\pm$ SD from 6 segments each in 6 volunteers), except (D) where points with peak power < 14 kW are from 3 volunteers. $IE^{4P} = -1$ would indicate perfect inversion. Smaller error bars imply more reliable inversion (smaller inter-subject or inter-segment variability).

Fig. 7.

Myocardial T_1 measured in vivo. Acquisition used optimal parameters except as specified.

a-b: T_1 (inter-segment mean $\pm$ SD) dependence on sequence parameters and the choice of post-processing as described in the Theory section. T_1^{4P} is most resistant to the deleterious influence of incomplete inversion (**a**) and readout-induced saturation (**b**). Connecting lines are drawn only to guide the eye and a small amount of horizontal jitter has been added for clarity. **c:** Summary of T_1^{4P} vs IE^{4P} . Data from the interseptal and anterosseptal segments acquired with optimal parameters are marked with a blue x; the mean $\pm$ SD of these points is marked in red. This is our most reliable measure of myocardial T_1 . Finally, to show that T_1^{4P} post-processing is robust, data from all segments in 177 acquisitions covering the wide range of parameters investigated for Fig. 6 are plotted with black dots. Grouping these points in intervals of 0.2 IE^{4P} , a black line shows the mean $\pm$ SD for each group, which vary modestly from those for the best measurements (in red).

Fig. 8.

Estimation of B_1^+ from ShMOLLI+IE data. Comparison of maps of B_1^+ per volt obtained by (a) fitting an HS8 pulse simulation to each pixel of the ShMOLLI+IE data in Fig. 3; and (b) actual flip-angle imaging (AFI) (34) B_1^+ sequence for reference. c: Scatter plot of corresponding pixels in a & b. d: In vivo B_1^+ for 6 mid-short-axis myocardial segments averaged over 3 volunteers (mean $\pm$ inter-subject SD).