Quantitative mapping of cardiac $T_1$ in vivo has proved valuable for clinical research at 1.5T & 3T. In a pilot study, we began adapting the ShMOLLI sequence for use at 7T. This work aims to achieve the achievable magnetization inversion and to maximise the accuracy of $T_1$ measurements in the heart at 7T. Hence, we further improve that "7T ShMOLLI" sequence and post-processing to address the particular challenges at 7T and present new phantom and in vivo experimental data to support these changes.

Methods: Fig. 1 shows "7T ShMOLLI" which has: optimised inversion pulses; FLASH readout instead of SSFP; and an 8 heart beat (hb) delay then 5 images to measure equilibrium magnetization $M_0$ and readout-induced saturation. In total, 12 images are recorded in a single breath hold. Validation used a phantom comprising 7 tubes of Gd-doped agar with a range of $T_1$s and $T_2$s, measured by standard IR-SE and SE experiments with TR=10s. Experiments in vivo used 6 male volunteers (21-29y, 59-82kg), a Magnetom 7T (Siemens), 16-element TEM transceiver array, 16×1kW RF amplifiers and ECG gating. Each volunteer was B$_1$-shimmere for a mid-short-axis slice and B$_0$-shimmere for an 8cm cube around the heart. Readouts were single shot FLASH with 2×GRAPPA, 144×100 matrix and 340×240×8mm FOV.

Theory: ShMOLLI processes pixels by a 3-parameter fit of signal at time $T_i$ after inversion
$$s(T_i) = A - B \exp(-T_i/T_1)$$
where $A$ and $B$ are constants and the time constant $T_1$ is corrected for small flip angles by
$$T_1^{\text{ShMOLLI}} = T_1 \left[ A/M_0 \right] = T_1 \left[ A/(B - A) \right] = T_1 \left[ B/(B - A) \right].$$

An inversion pulse is characterised by its inversion efficiency $IE = M_s(T_1 = 0)/M_0$, which varies between -1 for perfect inversion, 0 for saturation and +1 for a 0° pulse. Eq. (2) assumes perfect inversion $IE = -1$ so $M_0 = s(T_1 = 0)/IE = B - A$. At 7T, this assumption does not hold and Eq (2) gives erroneous $T_1$s that seem to vary with $IE$. A computationally efficient fit fixes images 1–7 to Eq (1) but then estimates $IE$ from image 8. In a few seconds this gives
$$T_1^{\text{ShMOLLI}} = T_1 \left[ A/M_0 \right] = T_1 \left[ s_8/\phi_{\text{image}}/A \right].$$

A slower, more detailed approach fits pixels from all 12 images $s(i)$ to a 4-parameter model that approximates the Bloch equation for $M_r(i)$. The 4 parameters are: relaxation time $T_1^{\text{param}}$, equilibrium signal $A = M_0$, inversion efficiency $IE$ and the fraction of $M_r$ preserved after each readout $F_i$. Each inversion or readout is taken to occur instantaneously at heart-rate dependent times recorded online and $M_i < 0$ whenever the phase difference $[\phi - \phi_{\text{image}}] > \pi/2$.

Analysis: Fitting of Eq (1-2) was online. Eq (3) and the 4-parameter fit were applied in Matlab. Poorly fitting points ($R^2 < 0.97$) were excluded from further analysis, which used manually drawn ROIs (in vivo, the 6 standard mid-short-axis segments).

Results: Phantom experiments (not shown) with 10–200V inversion pulses and 5° nominal readout revealed that $T_1^{\text{ShMOLLI}}$ seriously underestimate $T_1$ when $IE$ is low, but that $T_2^{\text{image}}$ is more resilient (for $IE < 0$, it has $-13\% - 5\%$) and $T_1^{\text{param}}$ is even better (errors $-6\% +4\%$).

Eq (3) corrects $T_2^{\text{image}}$ accurately for low flip-angle readouts, but under-estimates for larger flip angles (for 5° readout, errors $-13\% - 5\%$). Meanwhile, Fig. 2 shows that $T_1^{\text{param}}$ is accurate for a wide range of flip angles (for 10° readout, errors $-6\% +4\%$). Phantom data also supported the following trends which are presented here using in vivo results due to space.

In vivo, inversion efficiency $IE$ was measured for BIR4, HS (sech $\beta = 0.067$) and HSN (sech $\beta = 0.01$ and $n=2,4,6,8$). An optimised HS8 pulse ($R = 10$, $T_1 = 90\text{ms}$, 210V peak) had the best and most consistent inversion with mean $IE = 0.825$ to -0.788 per segment. There was no significant effect of readout flip angles $230°$ on $IE$ or $T_1^{\text{image}}$. Fig. 3 summarises 117 in vivo acquisitions, demonstrating that the 4-parameter fit practically eliminates any influence of $IE$ on $T_1^{\text{param}}$. Analysis with $T_1^{\text{image}}$ (not shown) gave mean values within 5% of Fig. 3, but with up to 70% larger standard deviation. Using the optimised inversion pulse, mean $T_1$s for the mid-short-axis myocardial segments are $T_1^{\text{param}} = 1892–1951\text{ms}$ and $T_1^{\text{image}} = 1846–1964\text{ms}$.

To test the RF power required for reliable $T_1$ mapping at 7T, measurements were repeated with only 150V inversion voltage (i.e. 7.2kW), which is more typical for 7T scanners. This gave mean $IE$ per segment: $-0.73\% - 0.51\%$ and $T_1^{\text{param}} = 1880–1976\text{ms}$. Finally, although Fig. 1 involves a long breath hold, simulations show that the 4-parameter fit is valid when the 5 extra images are acquired before the first inversion. This would reduce "7T ShMOLLI" to a 144b breath hold in future.

Conclusions: Accounting for imperfect inversion at 7T with 4-parameter fitting and using optimised pulse sequence parameters, the myocardial $T_1$s at 7T was measured to be $1926 \pm 22\text{ms}$. Other measurements fitting with Eq (3) or using only 150V (7.2kW) for inversion did not significantly alter this $T_1$ value, but did increase the variability.