Myocardial T1 Mapping with Phase-Sensitive Motion Correction and Inversion Recovery Fitting

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Introduction Myocardial T1 mapping enables the estimation of extracellular volume fraction as a means for the detection and quantification of diffuse myocardial fibrosis [1]. Myocardial T1 relaxation can be reliably estimated using the modified Look-Locker Inversion Recovery (MOLLI) sequence that samples the magnetization recovery after a non-selective inversion recovery (IR) preparation pulse [2]. This MOLLI acquisition is typically performed within a single breath-hold across several heart-beats using cardiac gating; however, myocardial motion often occurs due to imperfect breath-hold or varying R-R interval, which degrades pixel-wise T1 estimation. A fully automated motion correction directly utilizing MOLLI images is very challenging due to significantly different image contrast (Figure 1). Therefore, we propose a novel motion correction and T1 mapping scheme for MOLLI images based on a phase sensitive (PS) image reconstruction. This approach utilizes the fact that the acquired MOLLI images with long TI have well recovered magnetization; therefore, its phase can be utilized to restore the signal polarity for the entire MOLLI series. In this way, the varying image contrast during MOLLI series can be removed and registering MOLLI frames becomes more robust. Moreover, the inversion recovery fitting on MOLLI signals with restored polarity is more efficient and could lead to lower residual errors. In vivo evaluation was performed on 17 patients.

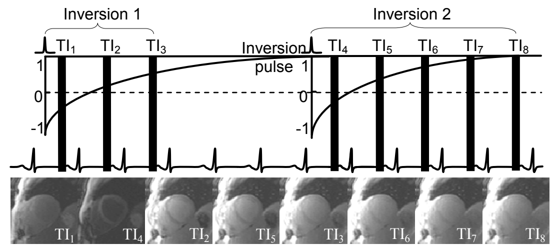

Figure 1. A MOLLI sequence scheme showing two passes of IR performed with increasing TI within one breath-hold. 8 images are acquired (3 from first IR and 5 from the second), as shown by the vertical bars. The magnitude MOLLI images show largely varying contrast between myocardium and blood pool.

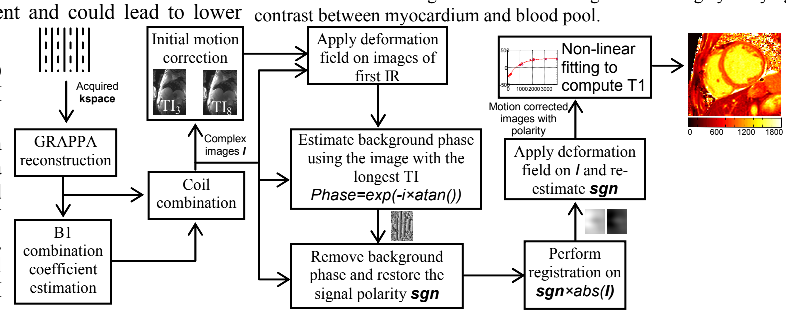

Figure 2. Flow chart of phase sensitive MOLLI mapping with motion correction and non-linear fitting.

Material and Methods ***In vivo study:*** 17 patients (8 males, 45.8±18.2 yrs) underwent MOLLI examinations using 1.5T scanners (MAGNETOM Avanto and Espree, Siemens AG Healthcare Sector, Erlangen, Germany). At least two slices (short axis and four chamber) were acquired for each subject both pre and post-contrast. Images at 8 TIs were acquired within a breath-hold: 3 after the first IR, 3 heart beats pause, and 5 after the second IR (Figure 1). Imaging parameters: SSFP, flip angle 35°, FOV 360×270mm^2, slice thickness 6 mm, minimum TI 120ms, increment 80ms, matrix 192×130, TR/TE 2.4/1.0 ms, 7/8 partial Fourier plus parallel imaging factor 2 with temporal resolution 140ms. A total of 63 MOLLI series (34/29 pre/post-contrast, 36/27 short/long axis) were acquired. ***Image reconstruction:*** As shown in Figure 2, the GRAPPA reconstruction was performed on acquired k-space data. The phased array combination coefficients were estimated using a Rank-1 eigen-analysis [3]. The SNR-optimal complex MOLLI images were computed by performing a B1-weighed array combination with the noise pre-whitening. ***Phase sensitive MOLLI motion correction:*** As the MOLLI images with the longest TI (>3000ms) can assure well recovered magnetization, their phase was removed from all MOLLI images on a pixel-by-pixel basis, and, as a result, the sign of the real part of the resultant image indicated the polarity of IR signal. While the phase is more spatially smooth, the initial misalignment between MOLLI images can still lower the accuracy of background phase removal. To correct this, an initial motion correction was applied between last images of IR experiments. For example, in the current protocol, the last image of the first IR was registered to the last image of the second IR. These two images have similar contrast due to large TIs, but could have noticeable mis-alignment as they were acquired at different times. The deformation fields were applied to other images of the first IR. The MOLLI magnitude signal was multiplied by the restored polarity (the real-part of complex signal is not used to avoid possible phase errors due to imperfect registration) and the contrast inversion between tissues was removed. A fast variational non-rigid registration algorithm [4] was applied here with localized cross correlation as the cost function. The estimated deformations were used to warp the original complex images and signal polarity was re-estimated once more on the motion corrected images. ***Phase sensitive MOLLI T1 mapping:*** Following the motion correction, pixel-wise T1 was estimated by performing only one non-linear least square fit to the IR exponential curve $A-B\times exp(-TI/T1^*)$ and $T1= T1^*\times(B/A-1)$ [1] on the polarity restored MOLLI signals. Compared to the magnitude based T1 mapping, the phase sensitive method has the clear advantage of much higher computational efficiency, because in order to detect the polarity of magnitude signal, the curve fitting has to be performed multiple times until the residual errors are minimized [1]; that is, the initial fit assumes that all data points are positive; the second fit inverts the first data point, the third inverts the first two points, and so on. Moreover, with the correct signal polarity, residual errors could be lower for the phase sensitive fitting because uncertainties of whether to invert points near signal nulling were removed. ***Quantitative validation of motion correction:*** All datasets were visually reviewed and classified into two categories: without and with motion. Two frames exhibiting motion were selected for each series with motion. Myocardium was manually delineated for these images. Four measures are computed: *Dice ratio* (the myocardium overlap ratio); *False positive/negative* (the percentage area of myocardium labeled/not-labeled in one frame but not-labeled/labeled in the other); *MBE* (the myocardium boundary errors, mean distance between endo/epi contours of two frames).

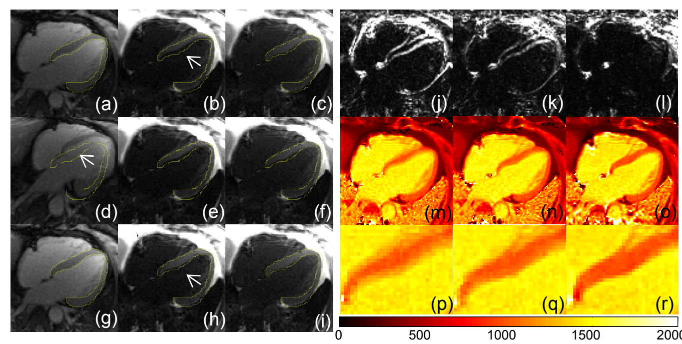

Figure 3. Example of PS MOLLI motion correction and T1 mapping. Three out of eight MOLLI images are shown here. (a-c) Original images showing noticeable motion. (d-f) Results by directly registration showing incorrect deformation. (g-i) Motion correction based on PS motion correction. (j-k) Fitting residuals of original magnitude MOLLI images and motion corrected magnitude images. (l) Residual errors of PS MOLLI fitting on motion corrected images. (m-o) T1 maps of original magnitude images, motion corrected magnitude images and PS MOLLI fitting. (p-r) Zoomed septum wall of T1 maps shows the improved quality by motion correction and phase sensitive fitting.

Table 1. The quantitative measures of motion correction.

	Dice		*FP*		*FN*		*MBE [mm]*	
	ori	*moco*	*ori*	*moco*	*ori*	*moco*	*ori*	*moco*
Mean	0.85	0.90	0.15	0.12	0.15	0.09	1.46	0.92
STD	0.05	0.02	0.06	0.05	0.07	0.04	0.05	0.02

Results 42 series were found to be free of noticeable motion and 21 with motion. A direct magnitude registration among MOLLI images often leads to unrealistic deformation (Figure 3d), which was found in 39 series among the whole cohort (62%). The proposed phase sensitive approach was very robust for drastic contrast changes. Table 1 summarizes the results, showing significant improvements of motion correction measures (P<0.05 for all 4 statistics). In this study, the phase sensitive T1 mapping was ~4x faster than magnitude mapping (~2s vs. ~8s). Figure 3 gives an example of T1 maps and fitting residual errors, illustrating the quality improvement of phase sensitive T1 mapping.

Conclusion Fully automated motion correction of MOLLI images can be achieved via the phase sensitive image reconstruction and non-rigid registration. The inversion recovery fitting on MOLLI signals with restored polarity is more efficient and could improve the map quality and lower residual errors.

References [1] Arheden H et al., Radiology 211:698-708 (1999) [2] Messroghli D et al., MRM 52:141-146 (2004) [3] Kellman P et al., MRM 47:372-383 (2002) [4] Chefd'hotel C et al., ISBI 753-756 (2002)