Automated respiratory motion correction approach for pixel-based analysis of first-pass MR myocardial perfusion studies

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INTRODUCTION Pixel-based analysis of first-pass MRI myocardial perfusion (FMP) studies has the advantage of using the full underlying spatial resolution of the images, but requires accurate image registration in the presence of respiratory motion to maintain the true spatial resolution. It has recently been shown that patient-adapted prospective navigator slice-tracking [1,2] can partially reduce 3D respiratory induced motion in FMP studies [3]. In this study, we show how the acquired navigators can be used retrospectively to further compensate the respiratory motion on a pixel-by-pixel level, in principle without user interaction. The proposed method relies on the basic assumption that respiratory motion affects the pixel signal similarly in image frames acquired in the same motion state (i.e. gating window), as determined by a preceding navigator. This, in combination with knowledge of the signal in neighbouring pixels, can be used to assign each pixel a signal value free of motion artifacts. To demonstrate its general performance, the proposed method was evaluated on first-pass myocardial perfusion images not corrected in real-time.

THEORY Let $\rho_i(r_0)$ denote the pixel signal at position r_0 in the i'th frame of the time-series, and let m_i denote the motion state during acquisition of the corresponding image. Suppose there are N distinct motion states over the total data acquisition and let m_n be the motion state such that $m_i = m_n$, and m_N the reference motion state. We then assume that $\rho_i(r_0)$ is a weighted sum of the pixel signals within its neighbourhood $N_d(r_0)$, depending on its displacement $x_n(r_0)$ in the n'th motion state (see Fig. 1). To a good approximation this can be stated as

$$\rho_i(r_0) \approx \sum_{\alpha} \Phi_{\alpha,i}\left[N_d(r_0)\right] w_{\alpha,n}(r_0), \qquad (1)$$

where $\Phi_{\alpha,i}$ is a set operation that depending on α calculates either the mean, minimum, or maximum signal value in $N_d(r_0)$ of the i'th image frame, and $w_{\alpha,n}(r_0)$ are weighting factors to be determined. The displacement $x_n(r_0)$ of r_0 in the n'th motion state is implicitly contained within the $w_{\alpha,n}(r_0)$'s. By ensuring that the number of frames acquired in the n'th motion state exceeds the number of terms (=3) in Eq. 1, the weighting factors can be determined using a least-squares fit. These weights provide an initial motion compensated estimate of $\rho_i(r_0)$ for each motion state. The validity of Eq. 1 is demonstrated in Fig. 2 (bottom), which shows the fitted signal for a subendocardial pixel in the end-expiratory and end-inspiratory motion states, respectively. Having calculated the weighting factors for all pixels and motion states, the displacement $x_n(r_0)$ of r_0 in all image frames acquired in the n'th motion state ($n \neq N$) can be estimated by solving the following minimization problem:

$$\underset{x_n(r_0)\in N_d(r_0)}{\arg\min} \sum_{\alpha} \left\| w_{\alpha,n}(r_0) - w_{\alpha,N}(r_0 + x_n(r_0)) \right\| \qquad (2)$$

Hence, for r_0 in the i'th image frame, the final motion compensated signal is $\rho_i(r_0) = \rho_i(r_0+x_n(r_0))$, where the ρ_i's are the initial signal estimates obtained from Eq. 1. To ensure consistency in the displacement of neighbouring pixels, it is necessary to constrain the above minimization process such that the displacement $x_n(r_0)$ is bounded by the corresponding displacement of the pixels lying immediately adjacent to r_0. We state here without proof that this can be formulated as a *linear programming problem*, which can be solved using standard algorithms. Fig. 3 shows the re-estimation of the end-inspiratory pixel signal from Fig. 2.

METHODS First-pass myocardial perfusion studies were conducted at rest on eight patients. All examinations were performed on a whole body 1.5T MR system (Gyroscan INTERA, Philips Medical Systems). Two short-axis slices were acquired in every other RR-interval for 120 consecutive cardiac cycles as described in detail in Ref. 3. A single navigator (N) located on the right diaphragm was applied in real-time between presaturation (S) and image acquisition (AQ), cf. Fig. 4, and contrast agent was administered in a dosage of 0.1 mmol/kg (Gd-DTPA). The acquired navigator was divided into a number of gating windows, as depicted in Fig. 2 (top), each corresponding to a separate motion state. A minimum of eight data points was required in each motion state, and the minimum size of each gating window was chosen to be 3 mm. In all cases, the end-expiratory gating window was chosen as the reference motion state. The radius *d* of the neighbourhoods $N_d(r_0)$ was set equal to the peak-to-peak amplitude of the navigator. The proposed motion compensation technique was applied to each pixel within all image sets. The efficiency of the compensation was assessed visually from CINE loops and by comparing pixel signals in corrected and non-corrected images.

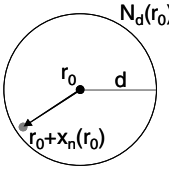

Figure 1. Illustration of the spatial parameters used in the theory section. The displacement of r_0 in the n'th motion state relative to the reference state is denoted $x_n(r_0)$. This displacement is expected to be smaller than some maximum value d, which defines the neighbourhood $N_d(r_0)$ of r_0.

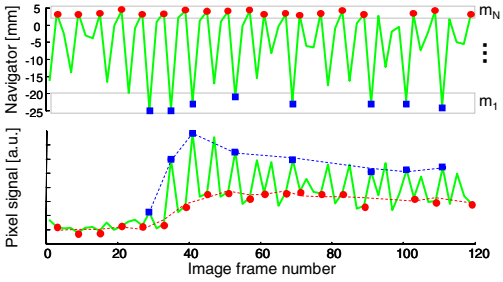

Figure 2. Typical signal in a subendocardial pixel (bottom) in the reference motion state, end-expiration (m_N), and in the end-inspiratory motion state (m_1). The motion states are shown from the navigator signal (top), and the result of fitting the pixel signals with Eq. 1 are shown as dashed lines (bottom).

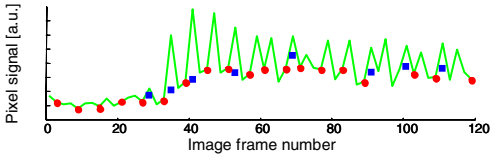

Figure 3. The pixel signal from figure 2 after correction based on Eq. 2 of all data points in the end-inspiratory motion state.

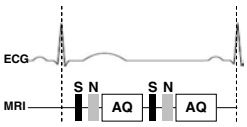

Figure 4. Timing diagram of the employed perfusion sequence.

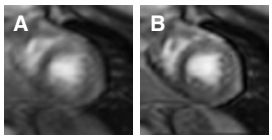

Figure 5. Comparison of the mean pixel intensities over time before (A) and after (B) motion correction.

RESULTS Experiments were successfully completed in all individuals. Inspection of the CINE loops showed only negligible traces of respiratory motion in the corrected images, which should be expected because the respiratory motion within each motion state is nearly stationary. Fig. 5 shows the mean pixel signal over time before and after motion correction of a typical data set. In this particular case, the CINE loop corresponding to the corrected images showed no trace of respiration, hence, the blurring seen in fig. 5A results from respiration. In general, corrected pixel signals demonstrated a least-squares like behaviour as expected from Eq. 1. In some image frames, however, the final estimation of the signal (Eq. 2) was slightly incorrect. This could be due to contaminations caused by other sources than respiration, or because a single navigator is inappropriate for characterizing the entire nature of respiration.

CONCLUSION The present study has shown that navigators can be used retrospectively to compensate respiratory motion on a pixel-by-pixel level in first-pass MR myocardial perfusion images. The algorithm can be applied to entire images, and therefore does not require any user interaction. A potential source of error in the current implementation is that a single navigator may not fully characterize the nature of respiration. In principle, this limitation can be overcome by using the navigator framework presented in Ref. 2, which potentially allows acquisition of 12 temporally and spatially distributed navigators for each image. Use of more navigators and the overall performance of the proposed method e.g. in pixel-based quantitative perfusion analysis needs to be addressed by further studies.

REFERENCES [1] Manke D, Magn. Res. Med. 2003;50:122–131. [2] Nehrke K, Magn. Res. Med. 2005;54:1130-1138. [3] Pedersen H, Proc. 13th ISMRM, p. 512.