Pipeline for Motion Correction in Dual Gated PET

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I. Introduction

Artifacts induced by respiratory and cardiac motion are a known problem in thoracic Positron Emission Tomography (PET). Elongated scanning times for PET in the range of minutes lead to unwanted image blurring due to constant patient motion. These motion artifacts may complicate clinical evaluation. Various approaches were proposed recently [1], [2] to overcome these limitations. The common motion correction scheme consists of three steps: 1) Gating, 2) Motion estimation, and 3) Motion compensation (e.g. averaging).

The aim of this work is to present an advanced motion correction pipeline for dual gating based on [3]. The pipeline is robust against noise and uses the whole statistic for the final image. We extend [3] by motion compensated reconstructions which allow a much finer dual gating for maximal reduction of motion artifacts. Further, mass-preserving motion estimation based on spline transformations is performed (VAMPIRE [2]).

II. Materials and Methods

A. General Pipeline

The incentive of the motion correction pipeline [3] is to reduce the $m \cdot n - 1$ motion estimation steps which are necessary in a naive approach, given an m-by-n dual gating (m respiratory and n cardiac gates). Instead of matching all $m \cdot n - 1$ gates to an assigned reference gate, only $m - 1$ respiratory matching steps with subsequent $n - 1$ cardiac matching steps are performed. It is assumed that respiratory and cardiac motion are independent to a certain degree. Hence, respiratory motion is estimated independently from cardiac motion. The transformation for a specific gate $\mathcal{T}$ with respect to the reference gate $\mathcal{R}_i^j$ is determined by concatenating the corresponding respiratory (y_i) and cardiac (y^j) transformation, i.e. $\mathcal{R}_i^j \approx \mathcal{T} \circ (y_i \circ y^j)$. The pipeline consists of the following steps: 1) Respiratory gating. 2) Classical reconstruction of respiratory gates (m images). 3) Respiratory motion estimation ($m - 1$ times). 4) Cardiac gating. 5) Respiratory motion compensated reconstruction of cardiac gates (n images). 6) Cardiac motion estimation ($n - 1$ times). 7) Dual gating. 8) Respiratory and cardiac motion compensated reconstruction (1 image). The scheme of the pipeline is illustrated in Fig. 1.

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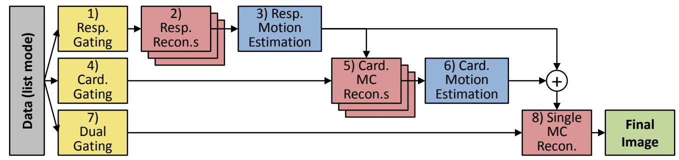


Fig. 1. Schematic illustration of the dual gating motion correction pipeline.

B. Patient data

20 minutes ^{18}F-FDG list mode scans of 13 patients with known coronary artery disease were performed. The list mode files were cropped to 3 minutes to have a clinically more realistic setting. All scans were performed on a Siemens Biograph™ Sensation 16 PET/CT scanner (Siemens Medical Solution) with a spatial resolution of around $6 - 7\,\mathrm{mm}$.

Dual gating was applied to the list mode data. An ECG signal was recorded for cardiac gating. A data driven method was used for the respiratory signal [4]. 10 respiratory ($m = 10$) and 10 cardiac ($n = 10$) gates were chosen to reduce the motion in each gate to a high extent.

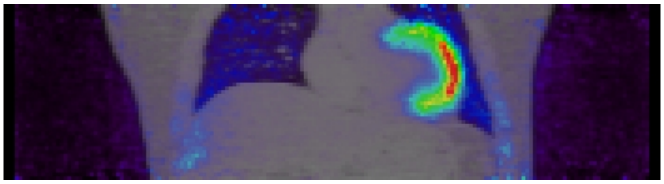
C. VAMPIRE - *Variational Algorithm for Mass-Preserving Image REgistration*

The mass-preserving image registration algorithm VAMPIRE is used in the motion estimation steps [2]. The template image $\mathcal{T} : \Omega \to \mathbb{R}$ is registered onto an assigned reference image $\mathcal{R} : \Omega \to \mathbb{R}$, where $\Omega \subset \mathbb{R}^3$ is the image domain. The objective is to find a transformation $y : \mathbb{R}^d \to \mathbb{R}^d$ that optimally aligns $\mathcal{T}$ and $\mathcal{R}$. In addition to the transformation of the template image $\mathcal{T}$, an additional multiplication by the volumetric change due to y is applied [2]. The following functional is minimized

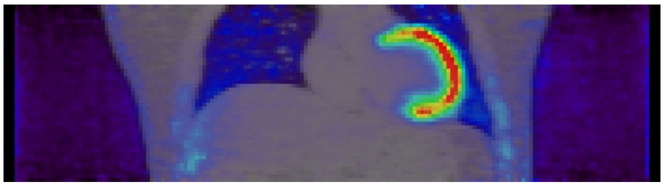
$$\min_y \frac{1}{2} ||(\mathcal{T} \circ y) \cdot \det(\nabla y) - \mathcal{R}||^2 + \mathcal{S}(y, M) \;, \qquad (1)$$

where $\mathcal{S}$ denotes the regularization functional and M a scalar weighting factor.

Transformation model. In [5] we found that both, spline transformations and displacement fields, yield comparable good results with mass-preserving motion estimation. Displacement fields guarantee diffeomorphic transformations but entail longer processing times. In this work we chose a spline transformation with regularization of the coefficients ($M = 5 \cdot 10^6$). For respiratory motion estimation (step 3)) the



(a) Without motion correction



(b) With motion correction

Fig. 2. Overlay of a coronal slice of the PET data and the attenuation map. (a): The resulting image without motion correction. (b): The resulting image with application of the proposed motion correction pipeline.

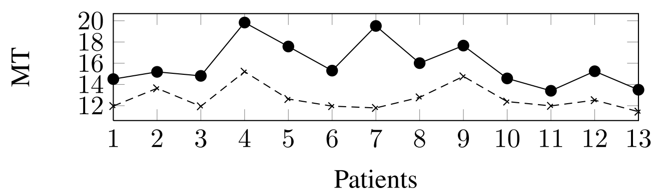


Fig. 3. The myocardial thickness (MT) before (solid line) and after (dashed line) motion correction is plotted for all 13 patients.

spacing of the coefficients was chosen larger with $50\,\mathrm{mm}$ to better model the rather locally rigid character. A finer spacing of $25\,\mathrm{mm}$ was applied for cardiac motion estimation (step 6)) to capture the high non-linearity.

D. Reconstruction

All reconstructions were performed with the 3D EM reconstruction software EMRECON [6], [7]. 5 iterations with 4 subsets were chosen. The output images are sampled with $175 \times 175 \times 47$, given an isotropic voxel size of $3.375\,\mathrm{mm}^3$.

Further, EMRECON allows a motion compensated reconstruction in the sense of [8] which was used for the cardiac reconstructions and for the final image.

III. RESULTS

The motion correction pipeline was successfully applied to 13 patient datasets. The results without and with motion correction can be seen in Fig. 2 for one patient. The highly reduced motion blurring of the heart contour is clearly visible.

The myocardial thickness (MT) was analyzed to quantify the reduction of motion. A Gaussian curve was fitted to a line profile through the left ventricle and the full width at half maximum (FWHM) was used as an indicator for the wall thickness. The results are shown in Fig. 3. For all patients we found a significant reduction of motion blur.

The Jacobian determinant of the estimated transformations is an indicator for the regularity of the transformation. The

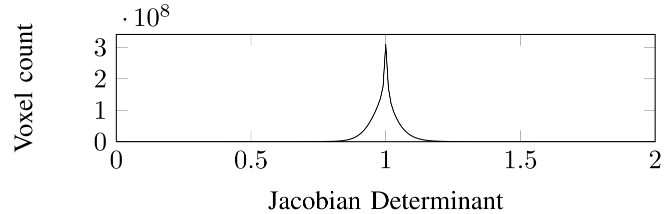


Fig. 4. Global histogram of the Jacobian determinant of all patients and all transformations. The histograms are added for all patients and for each patient the histograms of all transformations are added again.

Jacobian is always positive and centered around 1 for all transformations, see Fig. 4. The global minimum is 0.29 and the global maximum is 2.0 which implicates diffeomorphisms.

IV. DISCUSSION AND CONCLUSION

The presented motion correction pipeline represents a general framework including fine dual gating, mass-preserving motion estimation and motion compensated reconstructions. Even for relatively short acquisition times of 3 minutes a fine dual gating of 10×10 (i.e. 100 dual gates) is applicable due to the motion compensated reconstruction, whereas in [2], [3] reconstructions of all 100 single gates would be needed. The pipeline makes the motion estimation robust against noise as each gate contains $\sim \frac{1}{m}$th for the respiratory case ($\sim \frac{1}{n}$th for the cardiac case) of the whole statistic and not $\sim \frac{1}{m \cdot n}$th as in [2].

For future work we plan to apply attenuation correction during reconstruction. Further, the approach needs to be validated quantitatively on software phantom data. As a fine subdivision is now possible for dual gating, the optimal number of dual gates needs to be explored.

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