In-vivo MR-derived non-rigid motion correction of simultaneously acquired PET

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Purpose: Positron emission tomography (PET) is an important tool in the field of oncology. The desired patient-individual treatment of cancer requires accurate lesion detection and quantification which demands a high PET image quality in terms of signal-to-noise ratio (SNR) and sharpness (e.g. small full width at half maximum (FWHM)). Motion-induced artifacts originating from long free-breathing PET acquisitions lead to image degradation. A simultaneous PET/MR acquisition offers the possibility to correct these motion artifacts by deriving an MR-based motion model to correct the PET data[1]. It is essential to acquire a 4D (3D + time) MR motion model under free-breathing conditions as fast and accurate as possible to keep additional sequence workload low in order to provide a clinical feasible setup with diagnostic MR image quality. We therefore propose a dynamic (free-breathing) 4D Cartesian Compressed Sensing (CS) acquisition[2] which incorporates a self-navigation approach[3] to retrospectively determine the concordant motion state of the acquired MR sample. Additionally, we emphasize the use of a Compressed Sensing Partial Subsampling (ESPReSSo)[4] which compacts the allowed sampling region to a smaller subset, resulting in a denser sampling of the high frequency components and thus increasing edge delineation while improving registration accuracy. This acquisition scheme raises the need for a CS reconstruction. The motion model is build up on a multiresolution non-rigid image registration and multilevel cubic B-spline transformation. This work aims to create a clinical feasible motion correction (MC) setup for improving the PET image quality by correcting them with an accurate and fast acquired 4D MR-derived motion model without blocking the MR for MC.

Theory: *Acquisition:* A 3D Cartesian k-space is randomly filled over time based on a variable-density Gaussian ESPReSSo subsampling mask $\boldsymbol{\Phi}$ yielding a multi-coil 4D k-space[2]. For each T_{Nav}, the MR sequence is constrained to acquire a series of $M_{y,Nav} \times M_{z,Nav}$ k-space center samples from which a navigator signal $s(t)$ representing a projection of the moving liver dome over time, can be extracted.

Reconstruction: After retrospective gating of the acquired samples into their corresponding motion state $i \in [1, N_G]$ based on the navigator signal $s(t)$ and a view sharing blending factor which allows for overlapping gates, all $N_S = \max_i \#\{\text{samples state } i\}$ samples within a gate from N_c receiver coils are gathered to form the 6D subsampled k-space tensor $\boldsymbol{\nu} \in \mathbb{C}^{N_y \times N_x \times N_z \times N_G \times N_S \times N_c}$. The decision for the most significant sample representing the motion state i is performed by means of a sparsifying transformation $\boldsymbol{\Psi}$ in the CS reconstruction. Hence, this scheme allows maximal sample utilization. The image $\underline{\rho}$ is reconstructed from the stacked k-space $\underline{\nu} = \mathrm{vec}(\boldsymbol{\nu})$ using a regularized FOCUSS algorithm[5]:

$$\text{find } \underline{\rho} = \boldsymbol{\Psi}\boldsymbol{W}\underline{q} \text{ to } \min_{\underline{q}} \left\|\underline{q}\right\|_2 \text{ s.t. } \left\|\underline{\nu} - \boldsymbol{\Phi}\boldsymbol{\mathcal{F}}\boldsymbol{\Psi}\boldsymbol{W}\underline{q}\right\|_2 < \varepsilon,\ \Omega_1\colon \underline{\varphi} = \arg\left(\underline{\rho}\right) \cap\ \Omega_2\colon \underline{\rho} = \boldsymbol{\Psi}^{-1}\boldsymbol{\mathcal{F}}^{-1}\left(\boldsymbol{\mathcal{F}}\boldsymbol{\Psi}\left(\left|\underline{\rho}\right| \circ \exp\left(i\underline{\varphi}\right)\right) \cdot (\boldsymbol{I} - \boldsymbol{\Phi}) + \boldsymbol{\Phi}\underline{\nu}\right)$$

where $\boldsymbol{\mathcal{F}}$ denotes the Fourier and $\boldsymbol{W}$ the affine scaling transformation of the sparse image $\underline{q}$. The two closed convex sets Ω_1, Ω_2 integrate the ESPReSSo subsampling by formulating a projection-onto-convex-sets operation. The motion model τ_i for motion state i is derived by minimizing the dissimilarity $-S_M$ defined as the Normalized Mutual Information between the image $\underline{\rho_i}$ and the end-expiratory state $\underline{\rho_1}$: $\hat{\tau}_i = \arg\min_{\tau_i} -S_M\left(\tau_i, \underline{\rho_i}, \underline{\rho_1}\right)$. The determined displacement field $\vartheta_i\left(\underline{p}\right) = \tau_i\left(\underline{p}\right) - \underline{p}$ for each spatial voxel $\underline{p}$ is used to align the gated PET images. The attenuation map of each gate can be directly extracted from the corresponding motion image $\underline{\rho_i}$ using a 1-point DIXON method[6] to ensure a perfect motion alignment and hence no additional MR attenuation map scan is needed.

Materials & Methods: Coronal *in-vivo* patient data was acquired for 5 patients (4 female, age 56±4 years) with suspected lung metastases on a 3T scanner (Biograph mMR, Siemens) using a 3D spoiled gradient-echo sequence with TE = 1.23 ms, TR = 2.6 ms, a matrix size = 256 × 256 × 72 and a total scan time of 180 s. 4 gates were used without blending factor. Gated PET data were reconstructed using an 3D-OSEM algorithm with 2 iterations, 21 subsets and 4 mm Gaussian filter into a 256 × 256 × 127 matrix, transformed to the end-expiratory gate according to the MR-derived motion fields and summed into a motion-corrected image. Prominent moving liver and lung lesions were evaluated by means of ROI and line profiles in the corrected, uncorrected and end-expiratory gated PET images. Extracted ROI values were SUVmax = mean over maximum intensities inside ROI, SNR = SUVmax/standard deviation inside ROI and contrast = SUVmax/mean intensity in surrounding of the ROI. The line profiles were drawn parallel to the dominating movement direction (Head-Feet direction) with extracted FWHM and slope steepness. All three data combinations were evaluated in terms of percentage improvement.

Table 1: Mean and standard deviation improvements of 3 examined lesions

	Corrected vs. Uncorrected	Corrected vs. Gated	Uncorrected vs. Gated
SUVmax	15.73% ± 11.61%	5.73% ± 12.66%	-8.29% ± 10.27%
SNR	46.51% ± 53.63%	45.89% ± 47.82%	5.39% ± 32.81%
Contrast	8.63% ± 16.67%	-1.11% ± 9.19%	-8.15% ± 7.93%
FWHM	7.01% ± 10.26%	-12.42% ± 25.54%	-21.57% ± 25.70%
Slope	77.08% ± 30.85%	23.95% ± 22.88%	-27.98% ± 19.13%

Results & Conclusions: As it can be seen from Table 1, the corrected PET images show an improvement for all evaluated metrics compared to the uncorrected images, whilst the gated images also yield better results than the uncorrected ones. The comparisons of the corrected data against the gated data indicate a small decrease for contrast and FWHM originating from transformation and alignment of the different motion states. Fig. 1 shows a comparison of the obtained PET image quality and substantiates the overall improvement for the proposed MC scheme. The MR scan time can be further reduced down to 60 s, still yielding an accurate motion model. In conclusion, this setup provides a clinical feasible PET motion correction from MR-derived data with a short 4D Cartesian MR acquisition, resulting in significant improvement in PET image quality.

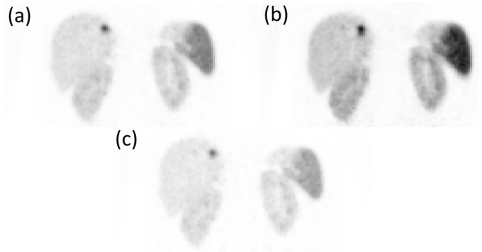

Fig. 1: Coronal PET images of the corrected (a), uncorrected (b) and end-expiratory gated (c) data.

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