Motion Correction Strategies for Integrated PET/MR

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ABSTRACT

Integrated whole-body PET/MR facilitates the implementation of a broad variety of respiratory motion correction strategies, taking advantage of the strengths of both modalities. The goal of this study was the quantitative evaluation with clinical data of different MR- and PET-data-based motion correction strategies for integrated PET/MR.

Methods: PET and MR data of 20 patients were simultaneously acquired for 10 min on a Biograph mMR PET/MR system after administration of FDG or DOTANOC. Respiratory traces recorded with a bellows were compared against MRI self-gating signals and signals extracted from PET raw data with the sensitivity method, by applying PCA or Laplacian Eigenmaps and by using a novel variation combining the former and one of the latter two. Gated sinograms and MR images were generated accordingly, followed by image registration to derive MR motion models. Corrected PET images were reconstructed by incorporating this information into the reconstruction. An optical flow algorithm was applied for PET-based motion correction. Gating and motion correction were evaluated by quantitative analysis of apparent tracer uptake, lesion volume, displacement, contrast and SNR.

Results: The correlation between bellows- and MR-based signals was 0.63±0.19 and 0.52±0.26 between MR and the sensitivity method. Depending on the PET raw-data compression, the average correlation between MR and PCA ranged from 0.25±0.30 to 0.58±0.33 and from 0.25±0.30 to 0.42±0.34, if Laplacian Eigenmaps were applied. By combining the sensitivity method and PCA or Laplacian Eigenmaps, the maximum average correlation with MR could be increased to 0.74±0.21 (PCA) and 0.70±0.19 (Laplacian Eigenmaps). The selection of the best PET-based signal for each patient yielded an average correlation of 0.80±0.13 with MR. Using the best PET-based respiratory signal for gating, mean tracer uptake increased by (17±19)% (gated), (13±10)% (MR-based motion correction) and by (18±15)% (PET-based motion
correction) compared to the static images. Lesion volumes were at (76±31)\% (gated), (83±18)\% (MR-based motion correction) and (74±22)\% (PET-based motion correction) of the sizes in the static images.

**Conclusion:** Respiratory traces extracted from MR- and PET-data are comparable with those based on external sensors. The proposed PET-driven gating method improved respiratory signals and overall stability. Consistent results of MR- and PET-based correction methods enable more flexible PET/MR scan protocols, while achieving higher PET image quality.

**Keywords:** PET/MR; multimodal imaging; respiratory gating; motion correction
INTRODUCTION

Physiological motion is a major source of deterioration of image quality in positron emission tomography (PET), leading to image blurring, rendering tumour uptake quantification less accurate and lesion volume delineation more difficult (1). Various methods for the reduction of motion artefacts in PET images were proposed, including gating (1), image-based approaches (2-4) and the incorporation of motion information into the reconstruction algorithm (5,6). Gating constitutes a trade-off of scan duration and image quality, because only a fraction of the acquired PET coincidences is taken into account for the reconstruction of individual gates. This leads to lower sensitivity, worse statistics and higher noise. Prolongation of examinations improves image quality, but adversely affects patient comfort and throughput. Despite this disadvantage, gating is still the most widely used respiratory motion compensation scheme.

For the correction of respiratory motion in PET, two types of information are essential, vector fields describing the motion within the body of the patient and a respiratory signal, which establishes the connection between motion model and PET data. The integration of whole-body PET with magnetic resonance imaging (MRI) and its introduction to the clinical routine (PET/MR) facilitates the implementation of a broad variety of motion-correction strategies, taking advantage of the respective strengths of both modalities. The respiratory signal can be obtained with external sensors, e.g. respiratory belts or cameras in combination with fiducial markers mounted to the chest of the patient (1,7,8), or extracted from PET data (9-11). In MR, navigator echoes are used to track the position and phase of objects of interest in the field of view (FOV) that are subject to respiratory motion, mostly the liver dome (12). With regard to the acquisition of motion vector fields, MR-based motion modelling techniques (13) appear to be the logical choice due to their analysis of physical deformation of anatomical structures for the detection of motion. However, other methods were published in recent years that promise to
derive the flow of activity directly from PET data (14). MR-based PET motion correction is widely assumed to be more robust due to the anatomical information and higher signal-to-noise ratio (SNR) of MR images, but is otherwise invasive and costly. Already demanding PET/MR workflows and scan protocols (15) would have to be extended for motion-modelling scans, whereas the acquisition of respiratory signals would block the subsystem with dedicated MR sequences throughout the entire PET examination. Alternatively, all clinical MR sequences would have to be interleaved with navigators, which might not be possible for any sequence with the sampling frequency required for respiratory gating or without inducing image artefacts. These factors might put further constraints on PET/MR protocols, limiting patient comfort and throughput. In contrast, PET-driven motion correction and its implementation might appear straightforward. However, it is still considered less accurate and reliable due to higher noise of PET data, subject to count rate statistics and tracer kinetics. None of the published studies of motion correction include a direct comparison of the two types of motion correction. The respective authors derived their results either from animal studies (16) or with simulated or phantom data (17-22). PET-driven motion correction was assessed with data of 14 patients (14), whereas MR-based motion correction was so far evaluated with data of 5 patients at maximum (23,24).

Therefore the goal of this study was the evaluation of different motion-correction strategies for integrated PET/MR. Within the scope of the present study, the authors propose a variation of methods for the extraction of respiratory signals from PET listmode data, testing the results on clinical data of 20 patients against MR- and sensor-based gating methods. Additionally, a rigorous comparative study of MR- and PET-driven motion-corrected PET image reconstruction was performed in terms of tracer uptake quantification, lesion volume definition and image quality with clinical data of 14 patients.
MATERIALS AND METHODS

Motion Correction Strategies

Respiratory Gating. As PET and possibly MR data have to be gated retrospectively, exact knowledge on the corresponding respiratory state is required at any time point during the examination. Within the scope of this study, this was achieved by the acquisition of respiratory signals according to five methods, i.e. the pressure-sensitive respiratory bellows (resp_bellows) that is shipped with the scanner by default, a prototype implementation of a self-gated MRI pulse sequence (resp_mr) (25,26), the PET-based sensitivity method (resp_sens) (9,10) and the application of dimensionality reduction techniques such as principal component analysis (PCA) or Laplacian Eigenmaps to the sinogram space (resp_pca, resp_le) (11,27,28). Prior to the execution of the PET-based methods, the listmode streams were divided into non-overlapping 200 ms time frames. The processing steps of resp_pca and resp_le are outlined in Figure 1. The number of sinogram planes was reduced to 127 by means of single-slice rebinning. All but the central 176 radial bins of each projection were cropped due to low counts and marginal respiratory motion. Seven different levels of radial compression and angular mashing were evaluated, ranging from 44 bins/9 projections to 1 bin/1 projection. PCA and Laplacian Eigenmaps were applied to the sinogram series using the MATLAB Toolbox for Dimensionality Reduction (29). A fifth method was proposed and evaluated by the authors, which builds on resp_pca(resp_le), but integrates resp_sens (resp_pca+sens, resp_le+sens) by omitting the normalisation in single-slice rebinning of sinogram bin values to the number of lines of response contributing to the corresponding plane. Thus, the axial sensitivity profile of the PET detector system was preserved in sinogram space. Apart from the normalisation, the processing regarding resp_pca+sens and resp_le+sens was identical to that of resp_pca and resp_le.
Calculation Of Motion Vectors. Sets of motion vector fields $M$ describing the inter-gate displacement of morphological structures are required for the warping of image volumes and acquired volumes of attenuation correction factors from the reference to the other respiratory states as defined by the gates. For MR-based motion correction (moco_mr), $M$ is calculated by acquisition of MR images at each gate and subsequent non-rigid registration. In the present study, the inverse motion vector fields $M^{-1}$ from gate to reference were generated by estimating for each voxel in the target image volume the corresponding voxel index in the reference after backward transformation to the reference. Special care was taken that spatial mismatch between an original voxel in the reference image and the corresponding voxel after the full warp cycle (from reference to gate to reference) was minimal, since such inconsistency would lead to a loss of resolution and adverse effects on image quantification in motion-corrected iterative PET image reconstruction.

In the absence of other imaging modalities, motion information can be derived from PET images using optical flow algorithms (14). The PET-driven method in this study (moco_pet) is based on a mass-conserving algorithm with an improved constraint, which assumes equal overall activity in each image (14).

Motion-Corrected Iterative PET Reconstruction. It is desirable to incorporate the two types of previously described information into the process of image reconstruction instead of manipulating reconstructed images. The most straightforward approach extends the conventional iterative ordered-subset expectation maximisation (OSEM) algorithm by integrating a warp function into the forward and backward projectors (Figure 2).

By combining the most suitable methods for each of these steps, a motion correction strategy could be optimised for specific clinical requirements.
**Patient Population**

The respiratory gating and motion correction methods presented above were applied to a total number of 20 patients (female, 11; male, 9) who had been referred to our department for the diagnosis and staging of malignant diseases (abdomen, 11; cardiac, 1; thorax, 8) using $^{18}$F-FDG (FDG) PET/CT or PET/MR or $^{68}$Ga-DOTANOC (DOTA) PET/MR (FDG, 18; DOTANOC, 2). They were (64±14) years of age and had a weight of (76±15) kg. The diseases include breast, liver and pancreatic cancer with additional lesions in e.g. lungs, lymph nodes, oesophagus and small intestine.

All patients gave written informed consent to participate in this study and, if only a PET/CT examination was clinically required and scheduled, undergo a second scan on the PET/MR. The approval of the institutional review board (project number, 2967/10) and the radiation protection authorities had been obtained. No additional radiotracers were injected after the first scan.

**Acquisition**

*Instrumentation.* All data in this study were acquired on a 3-T avalanche-photodiode-based integrated clinical PET/MR system (Biograph mMR, Siemens Healthcare, Germany; VB18P). The PET component of this tomograph is located between the gradient and the body coils. It covers axial and transaxial FOVs of 25.8 cm and 59.4 cm, respectively. Each detector block contains a matrix of 8 x 8 LSO crystals with a size of 4 x 4 x 20 mm$^3$, which are read out by 3 x 3 avalanche photodiodes. PET spatial resolution was measured to be 4.3 mm near and the PET sensitivity 15.0 kcps/MBq in the FOV centre ($^{30,31}$). Apart from the standard body coil and spine array coils, a flexible 6-element surface coil (Body Matrix Coil, Siemens Healthcare, Germany) was used for the MR measurements.
**Imaging Protocol.** Patients were injected with (339±63) MBq of FDG or 93 MBq/122 MBq of DOTA and scanned (136±24) min (FDG) or 66 min/52 min (DOTA) after injection. Prior to a scan, the patient was positioned on the scanner bed with the arms beside the torso and the respiratory bellows attached between costal arch and sternum to improve the signal for abdominal or thoracic breathing. One or two of the aforementioned body array coils were placed on abdomen or thorax, depending on the size of the patient and the body region to be examined as required by the indication. This region was then centred in the FOV of the scanner. A prototypical T1-weighted radial stack-of-stars spoiled 3D gradient echo sequence with fat suppression (sagittal slab orientation; FOV, 400 x 400 x 360 mm³; spatial resolution, 1.65 x 1.65 x 5 mm³; matrix, 256 pixels; 72 slices; 61% slice resolution; 5/8 partial Fourier) was employed for the subsequent derivation and calculation of MR-based respiratory signals and motion vectors fields. After generation of an attenuation map (µ-map) with a 2-point Dixon 3D volumetric interpolated breath-hold T1-weighted MRI sequence (TR/TE₁, 3.6 ms/1.23 ms; FOV, 500 x 399 mm²; voxel size, 4.1 x 2.6 x 3.1 mm³; 128 slices; flip angle, 10.0°; TA, 19 s), for which patients were given commands to hold their breath at end-expiration, PET listmode and radial MR data were simultaneously acquired for 10 minutes.

**Processing**

**Respiratory Signals And PET Gating.** All recorded respiratory traces were resampled to 10 Hz and then normalised for correlation analysis according to the following equation, where \( y(t) \) is the signal height at a time point \( t \) and \( \mu(Y) \) and \( \sigma(Y) \) the average height and corresponding standard deviation of the entire respiratory trace \( Y \):
\[ y_{\text{norm}}(t) = \frac{y(t) - \mu(Y)}{\sigma(Y)} \]  

(Eq. 1)

Listmode events were binned into sets of gated sinograms by means of variable-amplitude-based gating (32) according to the produced respiratory signals. The number of gates was set to 5, which constituted a good compromise between captured respiratory motion and noise in the reconstructed PET images and is identified as appropriate in the literature (33). Sinograms comprised of PET data from the entire scan (static) were also generated for each patient.

**Motion-Field Estimation.** Radial MR readouts were partitioned according to resp_mr, following the steps previously outlined for PET gating, and corresponding gated MR images reconstructed. A non-rigid registration algorithm proposed for lung imaging (34) was applied for the calculation of motion vectors. The gate comprising data recorded at end-expiration was selected as reference, since it is least affected by intra-gate motion under the current gating scheme.

**Attenuation Correction.** Patient µ-maps were created by segmenting the fat and water images generated by the Dixon MRI sequence. The original µ-maps with four tissue classes (35) were then utilised for the reconstruction of images that formed the basis of the respiratory gating analysis. For the reconstruction of PET images at end-inspiration, motion vector fields were applied to deform the µ-maps to this state. For the evaluation of motion correction, all voxels in the µ-maps classified as fat were assigned the attenuation coefficient of soft tissue and morphological structures in the lungs at or close to the location of lesions removed. This measure was taken to isolate the outcomes of motion correction from adverse effects caused by mismatch of attenuation and emission data at tissue boundaries. Truncation of arms in µ-maps due to the limited FOV of the MR subsystem was compensated by maximum-likelihood reconstruction of
attenuation and activity (36). For the reconstruction of attenuation-corrected PET images at end-inspiration, motion vector fields were additionally applied to deform the µ-maps from the reference to this state.

**PET Image Reconstruction.** Static, gated (reference state, gate 1, end-expiration; gate 5, end-inspiration) and motion-corrected (reference state, end-expiration) PET images were reconstructed from all previously described sinograms/sets of sinograms using the corresponding original and modified µ-maps and following the clinical standard in our department (OSEM 3D, 3 iterations, 21 subsets, 172 x 172 matrix, 4.0 mm Gaussian post-reconstruction filter) (31). In addition to attenuation, data were also normalised and corrected for dead time, scatter, decay, frame length and randoms.

**Analysis**

Respiratory signals of all 20 study patients were evaluated. Of these 20 patients, only those with moving lesions that exhibited tracer uptake sufficient for unambiguous segmentation in all static, gated and motion-corrected images were included for subsequent image-based analysis of respiratory gating and motion correction, resulting in 14 patient data sets with a total number of 27 lesions. For the validation of respiratory gating, only one lesion was considered per patient to avoid bias.

**Respiratory Signals.** The processed respiratory traces were evaluated qualitatively and by calculating the Pearson’s correlation coefficients of resp_mr with all other methods.

**Image Quantification.** The effects of gating and motion correction on reconstructed images were analysed in terms of apparent tracer uptake concentration in suspected tumour lesions and background tissue. For this purpose, isocontour volumes of interest (VOI) were segmented for each lesion individually by employing a region-growing algorithm (isocontour threshold, 50%).
Apart from the maximum activity concentration, the average activity concentration $A$ in each isocontour VOI was calculated, as well as standard deviation $\sigma$, contrast $C$ and signal-to-noise ratio $SNR$. The latter were defined similar to (37):

$$C = \frac{A_{\text{lesion}} - A_{\text{background}}}{A_{\text{background}}}$$  \hspace{1cm} (Eq. 2)

$$SNR = \frac{A_{\text{lesion}} - A_{\text{background}}}{\sigma_{\text{background}}}$$  \hspace{1cm} (Eq. 3)

For the analysis of background tissue, one VOI per patient was defined in the liver under the constraint not to include edge voxels or focal tracer uptake with a size as large as possible, but not larger than $11 \times 11 \times 21$ voxels ($4.6 \times 4.6 \times 4.3$ cm$^3$). Image noise was then expressed as relative standard deviation of all voxel values in the background VOI.

**Lesion Displacement And Volumes.** The position of an isocontour VOI along the craniocaudal axis was computed as its centre of mass as previously published (38). Lesion displacement $\Delta z$ was measured for the evaluation of respiratory gating as the absolute difference between the positions in gate 1 and gate 5. Lesion volume $V$ was defined as the total volume of all voxels within the isocontour VOI in the reference gate.

**Statistical Analysis.** The two-sided paired Wilcoxon signed-rank test was performed to determine statistical significance of the results. Differences, for which p-values of equal to or more than 0.05 were calculated, were regarded as statistically insignificant.

**RESULTS**

**Respiratory Signals**
For all 20 patients, respiratory traces could be successfully generated (Figure 3). The positions of inhalation peaks and overall noise of resp_bellows and resp_mr were visually well comparable and consistent. However, due to the limited recorded range of signal heights as currently implemented on the PET/MR scanner, inhalation peaks in resp_bellows were frequently cut off. Therefore, resp_mr was chosen as the reference signal.

The average coefficients of correlation between resp_bellows and resp_mr were 0.63±0.19 (max, 0.94; min, 0.13) and between resp_sens and resp_mr 0.52±0.26 (max, 0.86; min, 0.01). Dimensionality reduction techniques yielded average correlations between 0.25±0.30 and 0.42±0.34 (resp_mr/resp_le) and between 0.25±0.30 and 0.58±0.33 (resp_mr/resp_pca), depending on the level of mashing and radial compression. The preservation of the sensitivity profile in the processed data increased the maximum average correlation of resp_le+sens with resp_mr to 0.70±0.19. The application of PCA to a sinogram space with 44 radial bins and three projections with the sensitivity profile preserved, which will be denoted with resp_pca443sens in the following, resulted in the highest average correlation of 0.74±0.21 (max, 0.93; min, 0.06) between resp_mr and any PET-driven extraction method. If the PET-based extraction method that presented the highest correlation with resp_mr was selected for each patient individually, which will be referred to as resp_bestpet in the remainder of the paper, an even higher correlation of 0.80±0.13 (max, 0.93; min, 0.40) was achieved on average for the entire patient population. The coefficient of correlation with resp_mr was higher than 0.8 for 15% (resp_bellows), 15% (resp_sens), 55% (resp_pca443sens) and 65% (resp_bestpet) of all 20 patients (Figure 4).

Complete information with regard to all levels of compression can be obtained from Table 1.

Based on these results, gated and motion-corrected images, the analysis of which will be presented in the subsequent sections, were reconstructed from gated sinograms created according to resp_bellows, resp_mr and resp_sens and resp_bestpet. Corresponding coefficients of
correlation with resp_mr for the group of patients included in the image-based analysis were 0.71±0.11 (resp_bellows), 0.55±0.20 (resp_sens) and 0.81±0.09 (resp_bestpet). Images based on resp_pca443sens (correlation with resp_mr, 0.80±0.10) were not evaluated separately due to the similarly high quality of resp_pca443sens and resp_bestpet regarding the included patients.

The quantitative analysis of gated images is summarised in Table 2, whereas an example is shown in Figure 5.

**Motion-Corrected Iterative PET Reconstruction**

Both methods of motion correction included in this study improved the visual impression of the reconstructed PET images (Figure 6). Size and location of morphological structures in gated and both of the motion-corrected images were consistent. In static images, the same structures appeared blurred and larger. However, image noise was significantly lower in moco_mr and moco_pet than in gated images, whereas the noise patterns in static and motion-corrected images were virtually indistinguishable. Complete quantitative results on motion correction can be found in Table 3.

Static images exhibited a background noise level of (14.7±2.5)% on average. Depending on the respiratory-signal source, this level could be maintained at between (14.3±2.4)% and (14.5±2.6)% (moco_mr) and at between (14.8±2.3)% and (15.0±2.4)% (moco_pet) in motion-corrected images, but was between (29.2±4.2)% and (29.9±4.5)% in the gated images.

For resp_bestpet, neither of the observed differences in tracer uptake in lesions between gated and moco_mr (maximum uptake, \( p = 0.140 \); mean uptake, \( p = 0.274 \)), between gated and moco_pet (maximum uptake, \( p = 0.990 \); mean uptake, \( p = 0.572 \)) or between moco_mr and moco_pet images (maximum uptake, \( p = 0.116 \); mean uptake, \( p = 0.153 \)) were significant.
The differences in lesion volumes (gating method, resp_bestpet) were not significant for gated and moco_mr ($p = 0.058$) and gated and moco_pet ($p = 0.909$) images. For moco_mr and moco_pet, a p-value of $p = 0.029$ was calculated.

In terms of contrast and SNR, moco_mr and moco_pet were not significantly different ($p \geq 0.167$) with resp_bestpet as gating method, whereas the application of any motion correction improved both figures significantly over gating ($p \leq 0.028$).

**DISCUSSION**

The goal of this study was the assessment of different MR- and PET-data-driven respiratory gating and motion correction methods.

With regard to respiratory signals, the quality of resp_bellows was artificially reduced by the limited signal range supported in the current implementation on the PET/MR scanner and the resulting signal cut-off. Although this should not affect amplitude-based gating in most cases, the true correlation of resp_mr and resp_bellows is expected to be higher than the $0.63\pm0.19$ in this study.

The comparably low average correlation of resp_sens with resp_mr of $0.52\pm0.26$ combined with a higher standard deviation across the patient population reveals its lower reliability. Büther et al (10) reported a correlation of 0.65 between their implementation of resp_sens and respiratory signals obtained with a video-based method. Whereas the injected activity doses were similar, patients included in the present study had more than 30% lower activity levels at scan start due to the average delay of 135 minutes instead of 60 minutes, which might explain the difference in correlation. Moreover, Büther et al restricted their study to cardiac scans, where an extended hot object, i.e. the heart, is centred in the FOV, which could be an additional advantage for resp_sens. Support for this can be found in (28). Thielemans et al observed a correlation of approximately
0.50 between resp_sens and a video-based method, if one third of the acquired counts was deliberately rejected for the simulation of lower doses.

Of the more complex PET methods evaluated in this paper, resp_pca yielded better results than resp_le. The authors’ hypothesis could be confirmed that preservation of the sensitivity profile followed by the application of PCA or Laplacian Eigenmaps as proposed above increased quality and robustness of PET-driven respiratory signal extraction, especially towards lower activities in the patient’s body or with less specific tracer uptakes. In general, the higher the quality of resp_le/resp_pca was, the lower the benefit of resp_le+sens/resp_pca+sens. However, integration of the sensitivity method facilitated higher levels of compression without sacrificing resulting signal quality, which means that information lost due to compression could be compensated for. This leads to the possibility of extracting respiratory signals from PET listmode data under more difficult conditions, e.g. lower injected doses, and reduction of processing time or required computing power.

As shown with resp_bestpet, the signal quality could be increased with patient-specific compression parameters. This observation hints at a sensitivity of data-driven respiratory signal extraction to a combination of breathing patterns, activity levels and tracer distribution, which are very individual. Hence, there is not one single level of sinogram compression that yields the best result for every patient examination. The comparably low minimum correlation coefficient of 0.4 of resp_bestpet with resp_mr could be treated as an outlier. If this data point is excluded, the lowest correlation will be 0.67. Furthermore, 75% of resp_bestpet signals provide correlation coefficients higher than 0.70 and, in 30% of the cases, they are even higher than 0.90. Consequently, resp_bestpet was confirmed to be a feasible approach, if both MR and PET data are available for a common period of time.
The correlation coefficients of approximately 0.8 for resp_pca/resp_le as reported by Thielemans et al (11,28) could only be achieved with resp_le+sens/resp_pca+sens in the present study. Possible reasons for this discrepancy include the used levels of sinogram compression and sampling rates of the respiratory signals, the latter of which were five times higher in this study. Moreover, the studies were not performed on the same scanner models, which means that the corresponding results might be subject to different scanners’ sensitivities and scatter fractions. The longer scan duration of 10 minutes in the present study compared to the 3 minutes of Thielemans et al might also play a role.

Quantitative image analysis confirmed the correlation analysis. The only significant differences in terms of lesion displacement were found between resp_bellows/resp_sens and the other gating methods under investigation and in terms of lesion volume between resp_bellows and resp_pca443sens/resp_bestpet, which supports the notion that resp_bellows and resp_sens produce slightly inferior respiratory signals.

Results of motion correction had to be validated with both static and gated images. Gated images represent the reference for tracer uptake quantification and lesion volume. This should be achieved with motion correction, albeit at the noise levels of static images. As expected, both motion correction methods fully recovered the static noise level, whereas the noise exhibited by gated images was significantly higher, reducing SNR and hence lesion detectability.

Inferiority of PET-based motion correction to that driven by MR could not be confirmed in this study. On the contrary, the only statistically significant difference between the two motion correction methods for any of the analysed image properties was found in lesion volumes, if sinograms were created using resp_bestpet. Moreover, moco_pet appears to perform marginally better than moco_mr in terms of tumour uptake and volume. This could be due to two reasons. Firstly, the application of MR-derived motion fields to PET data constitutes an indirect approach,
whereas optical flow allows the direct calculation of motion information from PET data. The result of the latter would be a better agreement of the motion-corrected images with the images of the reference gate. Subsequently, the quality of the motion-corrected image would increase with the image quality of the gated reference. Secondly, MR motion fields were created by image warping using tri-linear interpolation. Otherwise, it would be difficult to connect information from multiple modalities. However, mass-preserved optical flow, which forms the basis of moco_pet, is superior to optical flow employing linear interpolation as shown by Dawood et al (14).

As supposed, SNR increased significantly over gating after the application of either motion correction method. This was similar for lesion contrast, for which only that in moco_pet images combined with resp_pca443sens did not change significantly. In general, these results highlight the consistency of both MR- and PET-driven motion correction methods.

Due to the limited number of real patient cases included in published studies, comparison is difficult. Würslin et al evaluated an MR-based motion correction of reconstructed PET images with data of five patients (23). Similar to our findings, they observed increases of 28%, 25% and 27% in motion-corrected relative to static images for maximum activity concentration in lesions, contrast and SNR, respectively. It has to be noted that within the scope of their study contrast was defined as ratio of maximum activity concentration in a lesion, instead of mean activity concentration in our case, over mean activity concentration in background tissue. Due to corresponding behaviour of mean and maximum tracer uptake in our study, this should not play an important role, however. When comparing motion-corrected and gated images, Würslin et al found only SNR to improve significantly, whereas maximum tracer uptake, volume and contrast were significantly worse. This contrasts with our observation and might hint at the advantages of motion correction incorporated into the reconstruction algorithm, where in the present study for
both evaluated methods contrast and SNR were measured to be significantly better than in gated images, but tracer uptake and volume were not significantly different (resp_bestpet). Petibon et al (24) reconstructed motion-corrected images of one liver case using MR-derived motion information in the reconstruction process. They reported comparable results, i.e. an increase in target-to-background ratio, which was defined identically to contrast as employed by Würslin, of between 22% and 45%, whereas apparent lesion volumes decreased by between 13% and 29%, depending on the lesion, if compared to static images. The results regarding PET-driven motion correction as published by Dawood et al (14) relate to myocardial thickness and blood-pool activity in the left ventricle in cardiac cases only and are therefore not comparable.

The consistency of MR- and PET-based motion correction methods as established in our study offers the operator of a PET/MR system a range of strategy choices tailored to a specific application. In oncological FDG studies, there will be the highest flexibility with regard to motion correction and scan protocols as the user can fully rely on PET-driven gating and correction methods. In more specialised cases with lower applied activities or less common radiotracers, moco_pet might have its limitations. For such purposes, motion models could be acquired with, e.g., radial MR sequences. During this time span of a few minutes, MR- and PET-derived respiratory signals could be compared and depending on the correlation decided, if PET respiratory signal quality was sufficient. If this was not the case, the system could automatically fall back to resp_mr or resp_bellows or a different external sensor. If in another scenario the user was not concerned with scan time, because the planned MR protocol was not demanding or because the clinician would be satisfied with the anatomical quality of the MR images acquired with the self-gating pulse sequence for motion modelling, moco_mr could be utilised in combination with resp_mr.
In terms of future work, a thorough investigation of the limitations of resp_pet and moco_pet with regard to count rate statistics should be performed. Moreover, the comparison of moco_mr and moco_pet should be extended to other more specialised applications, e.g. cardiac PET imaging. Thirdly, development of an algorithm to automatically select the optimal combination of radial bins and projections for resp_pca and subsequent quality assessment without additional MR information would be desirable.

**CONCLUSION**

Respiratory traces extracted from PET data are comparable with MR-derived signals and those based on external sensors. With the proposed PET-driven gating method, a higher quality of respiratory traces was achieved and the overall stability improved. Improvements of image quality, tracer uptake quantification and lesion volume delineation achieved with MR- and PET-based respiratory motion correction methods were consistent in evaluated oncological standard examinations, allowing for more flexible PET/MR scan protocols that employ solely PET-driven monitoring and correction. In lower-dose regimes, results of motion correction could be enhanced by the addition of external sensors or motion models derived from MR sequences.
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REFERENCES


FIGURES

FIGURE 1. Schematic of the workflow for the extraction of respiratory signals from PET data employing dimensionality reduction techniques.
FIGURE 2. Schematic of the workflow for incorporation of motion information into the OSEM reconstruction algorithm.
FIGURE 3. Respiratory traces of patient 8 from a 60-s mid-scan window. Correlation coefficients were 0.71 (resp_mr/resp_bellows), 0.79 (resp_mr/resp_sens) and 0.92 (resp_mr/resp_pca443sens). resp_bellows is cut-off above a certain signal height, whereas resp_sens appears noisiest. Visual impression confirms the high correlation between resp_mr and resp_pca443sens.
FIGURE 4. Cumulative frequency histogram of the coefficients of correlation of resp_bellows, resp_sens, resp_pca443sens and resp_bestpet with resp_mr as reference. The correlation coefficient is higher than 0.6 for 75% (resp_bellows), 40% (resp_sens), 90% (resp_pca443sens) and 95% (resp_bestpet) of all 20 patients.
FIGURE 5. Coronal slices through a lesion of patient 8, in images (gate 1) gated according to (A) resp_bellows, (B) resp_mr, (C) resp_sens and (D) resp_pca443sens. Visual differences in terms of lesion volume and location between (B), (C) or (D) are negligible. Compared to the other images, the lesion appears to be slightly shifted in (A), possibly indicating a higher degree of intra-gate motion.
FIGURE 6. Coronal slices through (A) static, (B) gated (gate 1), (C) moco_mr and (D) moco_pet images of patient 4. Gating was performed in all cases according to resp_pca446sens. Motion blurring of lesions (liver, arrows) was significantly reduced by gating and motion correction, leading to smaller volumes and higher apparent uptake. Whereas the reduced number of counts is obvious in (B), noise patterns in (A), (C) and (D) are consistent.
TABLE 1

Dependence of PET-based respiratory trace quality on sinogram-space compression.

<table>
<thead>
<tr>
<th>Radial Bins</th>
<th>Projections</th>
<th>Correlation Coefficients</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>resp_le</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td>0.25±0.30</td>
</tr>
<tr>
<td>11</td>
<td>1</td>
<td>0.30±0.32</td>
</tr>
<tr>
<td>22</td>
<td>1</td>
<td>0.34±0.34</td>
</tr>
<tr>
<td>44</td>
<td>1</td>
<td>0.31±0.32</td>
</tr>
<tr>
<td>44</td>
<td>3</td>
<td>0.42±0.34</td>
</tr>
<tr>
<td>44</td>
<td>6</td>
<td>0.39±0.32</td>
</tr>
<tr>
<td>44</td>
<td>9</td>
<td>0.37±0.30</td>
</tr>
</tbody>
</table>

Reference: resp_mr

Shown values are averaged over the entire patient population (mean±standard deviation).
<table>
<thead>
<tr>
<th>Lesions</th>
<th>Max. activity concentration</th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Static (kBq/mL)</td>
<td>resp_bellows</td>
<td>resp_mr</td>
<td>resp_sens</td>
<td>resp_bestpet</td>
</tr>
<tr>
<td></td>
<td>21.7±17.5</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Gate 1 (% of static)</td>
<td>-</td>
<td>113.5±11.6</td>
<td>116.9±15.5</td>
<td>110.4±9.1</td>
<td>117.1±15.6</td>
</tr>
<tr>
<td>Mean activity concentration</td>
<td>Static (kBq/mL)</td>
<td>14.4±11.6</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Gate 1 (% of static)</td>
<td>-</td>
<td>111.3±10.1</td>
<td>114.8±13.8</td>
<td>109.8±7.9</td>
<td>114.3±11.4</td>
</tr>
<tr>
<td>Volume</td>
<td>Static ($10^3$ mm$^3$)</td>
<td>6.5±8.9</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Gate 1 (% of static)</td>
<td>-</td>
<td>80.1±23.5</td>
<td>76.6±24.6</td>
<td>81.3±19.2</td>
<td>73.5±30.3</td>
</tr>
<tr>
<td>Displacement</td>
<td>Gate 1 - Gate 5 (mm)</td>
<td>-</td>
<td>4.4±2.6</td>
<td>5.2±3.0</td>
<td>3.8±3.3</td>
</tr>
</tbody>
</table>
### TABLE 3
Summary of results from the image-based analysis of motion correction methods.

<table>
<thead>
<tr>
<th>Lesions</th>
<th>static</th>
<th>resp_bellows</th>
<th>resp_mr</th>
<th>resp_sens</th>
<th>resp_bestpet</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max. activity concentration</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Static (kBq/mL)</td>
<td>18.6±11.9</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Gate 1 (% of static)</td>
<td>119.9±16.8</td>
<td>119.3±15.5</td>
<td>114.7±15.0</td>
<td>116.4±18.3</td>
<td>-</td>
</tr>
<tr>
<td>moco_mr (% of static)</td>
<td>109.2±9.4</td>
<td>112.2±10.2</td>
<td>104.1±5.9</td>
<td>111.1±8.3</td>
<td>-</td>
</tr>
<tr>
<td>moco_pet (% of static)</td>
<td>110.4±15.6</td>
<td>114.1±16.8</td>
<td>110.7±15.6</td>
<td>116.8±16.5</td>
<td>-</td>
</tr>
<tr>
<td>Mean activity concentration</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Static (kBq/mL)</td>
<td>12.5±8.0</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Gate 1 (% of static)</td>
<td>120.5±17.8</td>
<td>119.6±15.5</td>
<td>115.5±17.0</td>
<td>116.5±19.0</td>
<td>-</td>
</tr>
<tr>
<td>moco_mr (% of static)</td>
<td>110.5±10.3</td>
<td>113.6±11.9</td>
<td>104.7±6.3</td>
<td>112.7±9.8</td>
<td>-</td>
</tr>
<tr>
<td>moco_pet (% of static)</td>
<td>110.3±15.6</td>
<td>115.3±14.7</td>
<td>111.2±13.7</td>
<td>117.7±15.4</td>
<td>-</td>
</tr>
<tr>
<td>Volume</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Static (10^3 mm³)</td>
<td>4.0±6.5</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Gate 1 (% of static)</td>
<td>69.5±25.9</td>
<td>70.3±25.1</td>
<td>74.1±23.1</td>
<td>75.8±30.6</td>
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<tr>
<td>moco_mr (% of static)</td>
<td>83.7±18.4</td>
<td>82.2±18.5</td>
<td>91.7±16.4</td>
<td>82.7±18.4</td>
<td>-</td>
</tr>
<tr>
<td>moco_pet (% of static)</td>
<td>84.7±21.1</td>
<td>77.3±19.5</td>
<td>82.6±20.1</td>
<td>74.0±21.7</td>
<td>-</td>
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<tr>
<td>Contrast</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Static (%)</td>
<td>47.3±31.2</td>
<td>-</td>
<td>-</td>
<td>-</td>
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<tr>
<td>Gate 1 (% of static)</td>
<td>108.0±92.2</td>
<td>97.6±100.5</td>
<td>109.3±53.6</td>
<td>113.3±59.4</td>
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<tr>
<td>moco_mr (% of static)</td>
<td>106.4±43.7</td>
<td>107.5±51.6</td>
<td>116.3±52.6</td>
<td>110.2±42.7</td>
<td>-</td>
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<tr>
<td>moco_pet (% of static)</td>
<td>112.1±31.9</td>
<td>112.1±18.9</td>
<td>104.0±9.2</td>
<td>114.8±25.0</td>
<td>-</td>
</tr>
<tr>
<td>SNR</td>
<td></td>
<td></td>
<td></td>
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<td></td>
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<tr>
<td>Static (%)</td>
<td>1,300±1,396</td>
<td>-</td>
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<td>-</td>
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<tr>
<td>Gate 1 (% of static)</td>
<td>65.9±61.7</td>
<td>60.3±66.4</td>
<td>65.7±41.0</td>
<td>71.3±40.8</td>
<td>-</td>
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<tr>
<td>moco_mr (% of static)</td>
<td>119.3±60.8</td>
<td>125.6±73.8</td>
<td>120.9±52.3</td>
<td>126.7±62.6</td>
<td>-</td>
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<tr>
<td>moco_pet (% of static)</td>
<td>119.6±35.2</td>
<td>125.9±30.6</td>
<td>114.3±21.1</td>
<td>132.8±40.1</td>
<td>-</td>
</tr>
</tbody>
</table>
Motion Correction Strategies for Integrated PET/MR

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