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MR-based Motion Correction for Quantitative PET in Simultaneous PET-MR Imaging

Yothin Rakvongthai, PhD¹ and Georges El Fakhri, PhD^{2,3}

¹Division of Nuclear Medicine, Department of Radiology, Faculty of Medicine, Chulalongkorn University, Bangkok 10330, Thailand. ²Gordon Center for Medical Imaging, Department of Radiology, Massachusetts General Hospital, Harvard Medical School, Boston, MA, 02114, USA.

Keywords

PET-MR; Respiratory motion; Cardiac motion; MR-based motion correction

1. Introduction

Patient motion degrades image quality and quantitation of positron emission tomography (PET) images, and is an obstacle to quantitative PET imaging. The blurring owing to motion leads to the underestimation of tracer uptake values^{1,2}, and the reduction of lesion detectability^{3–6}, especially for small lesions. For whole-body PET scanners, the intrinsic spatial resolution is approximately 4 mm^{7,8}, which could not be achieved in clinical thoracic and abdomen studies due to unavoidable patient motion. In particular, during respiration, the diaphragm and the liver moves up to 28 mm and 17 mm⁹, respectively, which severely affects detection of lesions in the liver dome^{3,4}. Moreover, the heart moves up to 9 mm during free breathing⁹. In addition to blurring, the motion creates discrepancy between the emission and attenuation data, which leads to confounding artifacts in PET images, and thus reducing PET quantitative accuracy.

There are several ways to generate PET images in the presence of motion. First, one can mistakenly ignore the motion by assuming that the object to be imaged has no motion, and reconstruct the PET image (this is called the no motion correction or the uncorrection method). There have been research studies that attempt to overcome the effects of motion. Tracking systems using external devices¹⁰ that link external movements to internal organs' trajectories has been proposed, but the errors are still substantial. PET data alone can also be used to estimate the motion; however, its accuracy is limited by the relatively low spatial resolution of PET, and it only works well for regions with high activity¹¹. Another widely-used scheme to deal with motion is the gating method, which freezes the cardiac and/or the respiratory motions. In this case, the motion trajectory is subdivided into phases and acquired PET data are binned into multiple frames or gates according a specific motion phase. Consequently, the PET image is reconstructed from one chosen reference frame to

³Corresponding author (elfakhri@pet.mgh.harvard.edu).

represent a motion-free PET image. Nevertheless, this strategy uses only a small fraction of the emission data for reconstructing PET image in each individual gate. This leads to a tradeoff of the image's signal-to-noise-ratio (SNR) and the total scan time; and therefore it is not optimal.

Simultaneous PET-magnetic resonance imaging (MRI or MR) offers a tool which can be used for correcting the motion in PET images by using high quality anatomical information from MR imaging. Unlike in PET-computed tomography (CT), motion-induced blurring can be alleviated in PET-MR without additional radiation exposure or loss of SNR while both emission data and data used for creating motion fields are acquired concurrently. The combination between PET and MR also allows us to take advantages of both modalities.

The MR-based PET motion correction consists of two parts: estimating deformation field or motion field from MR data, and creating a motion-free PET images using the estimated deformation field. This so-called motion field or deformation field is a vector fields representing the displacement of all individual voxels from every motion phases to a chosen reference phase. To perform motion estimation from MRI, the motion fields are extracted from measured MR data which are acquired under MRI acquisition protocols specially designed for fast dynamic imaging. In addition to PET motion correction, motion estimation is also used to generate the time-dependent attenuation maps to be used for PET image reconstruction. To create the motion-corrected PET image, the first approach uses data from all gates by reconstructing PET images of all frames, and transforming them into the reference frame. The resulted transformed images are averaged to have the motion-corrected PET image. Instead of transforming PET images in the post-reconstruction step, another approach incorporates the transformation into the reconstruction system model, and obtains the motion-corrected PET image in a single reconstruction framework.

2. MR-based Motion Measurement and Motion Field Estimation

Two major causes of motion-induced blurring artifacts are the respiratory motion and the cardiac motion. This section discusses how to measure the respiratory and the cardiac motions using MRI to assist PET motion correction in thoracic/abdomen and cardiac PET-MR imaging.^{12,13}

A dedicated MRI pulse sequence called NAV-TrueFISP was developed to measure the respiratory motion for lower abdomen PET-MR imaging.¹⁴ It takes advantage of specific contrast of single-slice steady-state free precession MRI acquisitions (TrueFISP) to produce anatomical landmarks in homogeneous tissues in the lower abdomen such as liver by combining TrueFISP with interleaving pencil-beam navigator echoes. Collected prior to each TrueFISP acquisition, these navigators track the lung-liver interface during the respiratory cycle. The resulted internal motion surrogates yield accurate binning of simultaneously acquired PET-MR data into respiratory phase, which is performed based on the navigator amplitude. As a result, gated MR image volumes can be used for computing the respiratory motion field. Another approach used a navigator-encapsulated golden angle radial FLASH pulse sequence to obtain time-dependent lung MR images in pulmonary PET-MR imaging.¹⁵ It should be mentioned that a patient study in abdomen, thoracic, and cardiac

imaging compared performance of several methods to acquire motion field in simultaneous PET-MR.¹⁶

For correcting motion in cardiac imaging, MR-based cardiac motion estimation involves two motion types: cardiac motion and respiratory motion. The motion field for these two types are measured separately, and are combined to create the motion field between any cardiac or respiratory phase to the reference phase.

The cardiac motion can be traced using tagged MR imaging technique, where series of parallel strips are created on the muscle tissue. These strips are called tags which are induced by periodic magnetization modulation due to specific radiofrequency (RF)/gradients series. A series of MR images is obtained in which the tags are visible and their deformation is used to estimate cardiac motion. The sequence is also navigated by a pencil-beam navigator, allowing the acquisition of tagged MR data only at end-expiration respiratory phase. Electrocardiogram (ECG) signals and navigators are acquired simultaneously for labeling each PET coincidence into a cardiac and respiratory phase. To speed up from regular tagging acquisition (10 min), an accelerated acquisition has been developed using advanced MR techniques such as compressed sensing.¹⁷ It should be noted that while tagged MR imaging technique has been successfully used in phantoms, abdomen imaging of rabbits and nonhuman primates, it is impractical for respiratory motion estimation in human studies because tag lines fade rapidly to capture the longer human breathing cycle.^{14,15}

Deformation fields from MRI is essential for PET motion correction. Accurate motion correction needs volumetric motion fields obtained from non-rigid registration of anatomical images acquired simultaneously with PET. To estimate the motion field from series of MR images volumes, B-spline non-rigid image registration, which is based on the sum of squared difference (SSD) or mutual information (MI)¹⁸, can be used. Denote an MR image volume at a given motion phase k by $\mathbf{f}(k, \mathbf{v})$. Given that the SSD is used, the motion fields are the solution to this minimization problem:¹⁴

$$\hat{\mathbf{g}}(k \rightarrow k', \mathbf{v}) = \underset{\mathbf{g}}{\operatorname{argmin}} \left[\frac{1}{N} \sum_{\mathbf{v}} \left(\mathbf{f}(k, \mathbf{g}(k \rightarrow k', \mathbf{v})) - \mathbf{f}(k', \mathbf{v}) \right)^2 + \beta R(\mathbf{g}(k \rightarrow k', \mathbf{v})) \right],$$

where $\mathbf{v}$ is the voxel position, N is the total number of voxels, β is a regularization parameter, and $R(\cdot)$ is a regularizer. The regularization is present to achieve stable and realistic solution because motion estimation is an ill-posed inverse problem.

3. Motion Correction in PET Reconstruction

The motion corrected PET image, which is the PET image corresponding a given reference frame or gate, can be obtained using two approaches, namely the reconstruct-transform-average (RTA)^{19–21} and the motion compensated image reconstruction (MCIR)^{22–26} methods. In the first approach, the PET image in each frame is reconstructed separately, and is then transformed back into the reference frame. All resulted transformed images are averaged to generate the motion-corrected PET image. Unlike the RTA method, the MCIR method includes the transformation in the PET system model, and reconstructs the motion-

corrected PET image in a single reconstruction framework. This section discusses these two PET motion correction approaches.

3.1 Reconstruct-Transform-Average (RTA) Approach

Once the deformation fields mapping PET image in the reference frame to all frames are obtained, the motion-corrected PET image in the reference frame can be reconstructed using the RTA method. To begin with, the PET system model for frame k is given by

$$\bar{y}_k = \mathbf{H}_k \mathbf{x}_k + \mathbf{s}_k + \mathbf{r}_k$$

where $\bar{y}_k$ is the mean PET sinogram in frame k , $\mathbf{x}_k$ is the PET image, $\mathbf{s}_k$ and $\mathbf{r}_k$ are the average scattered and random counts in the same frame. The system matrix $\mathbf{H}_k$ for frame k incorporates the detector sensitivity, detector blurring, frame-dependent attenuation factors, forward projection operator, and the warping operator transforming the reference frame to the k th frame. Mathematically, the system matrix can be written as

$$\mathbf{H}_k = \mathbf{N} \mathbf{B} \mathbf{A}_k \mathbf{G},$$

where $\mathbf{N}$ is a diagonal matrix accounting for the detector normalization factors, $\mathbf{B}$ models the detector blurring effects in the sinogram domain, $\mathbf{A}_k$ is a diagonal matrix representing the attenuation correction factors for each frame, $\mathbf{G}$ is the forward-projection operator whose (i,j) -th element represents the probability that an emission in voxel j is detected in detector bin i . A standard reconstruction method such as the ordered-subset expectation maximization (OSEM) algorithm can be used to obtain the estimated PET image in frame k , $\hat{\mathbf{x}}_k$. The final motion-corrected PET image in the RTA method is then given by

$$\hat{\mathbf{x}}_{\text{RTA}} = \frac{1}{K} \sum_k \mathbf{M}_k^{-1} \hat{\mathbf{x}}_k,$$

where, $\mathbf{M}_k$ is an interpolation matrix accounting for warping operator for transformation from the reference frame to the k th frame, and K is the total number of motion frames.

Another variant of this RTA method has been proposed²⁷ where each frame in the average step is weighted by the reciprocal of the relative amplitude change in that frame.

3.2 Motion-Compensated Image Reconstruction (MCIR) Approach

Even though there are studies^{27,28} showing that motion correction with the RTA method yields improvement in terms of image quality in comparison with the uncorrected method, the RTA method involves the transformation in post-reconstruction step, and therefore is sub-optimal. A more sophisticated method for applying motion correction in PET modifies the image reconstruction process. This method is referred to as the motion-compensated image reconstruction (MCIR). Unlike the RTA method where the transformation is performed after reconstruction, the MCIR method incorporates the transformation in the system model, and reconstruct the motion-corrected PET image in the reference frame in a

single reconstruction framework^{14,15,25,29–32}. In this case, the PET system model can be written as

$$\bar{y}_k = \mathbf{P}_k \mathbf{x} + \mathbf{s}_k + \mathbf{r}_k,$$

where $\mathbf{x}$ is the PET image in the reference frame to be reconstructed. In addition to all physical effects, the system matrix $\mathbf{P}_k$ for frame k in MCIR also incorporates the warping operator transforming the reference frame to the k th frame. The system matrix in MCIR can be written as

$$\mathbf{P}_k = \mathbf{N} \mathbf{B} \mathbf{A}_k \mathbf{G} \mathbf{M}_k.$$

To reconstruct the motion-corrected image in the MCIR framework, the widely-used OSEM algorithm can also be extended, and its update equation at iteration i is given by¹⁴

$$\mathbf{x}^{[i+1]} = \frac{\mathbf{x}^{[i]}}{\sum_k w_k \mathbf{M}_k^T \mathbf{G}^T \mathbf{A}_k \mathbf{B} \mathbf{1}} \times \sum_k \mathbf{M}_k^T \mathbf{G}^T \mathbf{B} \frac{\mathbf{y}_k}{\mathbf{B} \mathbf{G} \mathbf{M}_k \mathbf{x}^{[i]} + (\mathbf{A}_k \mathbf{N})^{-1} (\mathbf{s} + \mathbf{r})},$$

where $\mathbf{1}$ is a column vector whose elements are all 1, $\mathbf{y}_k$ is the measured PET sinogram in frame k , and w_k is the relative duration of frame k . It is assumed that the effects of random and scatter coincidences do not change with the motion frames; therefore the subscripts of $\mathbf{s}$ and $\mathbf{r}$ are omitted. This iterative approach yields the motion-corrected PET image in the MCIR framework, $\hat{\mathbf{x}}_{\text{MCIR}}$. It should be noted that in addition to the OSEM, the maximum a posteriori (MAP)^{33,34} reconstruction can be also used for the MCIR framework. A MAP reconstruction with quadratic penalty function as a prior together with a preconditioned conjugate gradient algorithm was proposed for motion correction in lung PET imaging with integrated PET-MR¹⁵. Figure 1 illustrates the diagram for the RTA and the MCIR methods for PET motion correction. Several phantom and patient studies^{35–38} compared both approaches based on the OSEM image reconstruction and regularized reconstruction, and reported that the MCIR approach yielded better motion-corrected images as compared with the RTA approach.

4. Impact of Motion Correction on Quantitative PET Using PET-MR

Applying MR-based motion correction in simultaneous PET-MR yields improved quantitative PET imaging. This section presents results of motion correction and its impact on quantitative PET using simultaneous PET-MR.

Cardiac imaging is one of several areas that benefits from advancement of PET motion correction in simultaneous PET/MR^{14,32}. It was reported³² that MR-based motion correction yielded an improvement in contrast recovery of 34–206% in comparison with no motion correction method, and in myocardial defect detectability of 115–136% and 62–235% as compared with the gating and no motion correction methods, respectively. In a plaque imaging study⁹, motion correction improved plaque detectability in terms of the channelized

Hotelling observer signal-to-noise ratio (CHO-SNR) by 105%–128% and plaque contrast by 30–71% as compared with no motion correction, and by 348% and 396% as compared with the gating and dual (cardiac-respiratory) gating methods. Figure 2 shows impact of motion correction in a cardiac lesion detection. Motion correction visually and quantitatively had better noise control while maintaining comparable contrast as compared with the gating method.

Motion correction also has impact on oncologic PET imaging. A quantitative oncologic PET-MR imaging framework involving MR-based motion correction and PSF compensation has been proposed.¹⁴ The study therein showed the improvement due to motion correction in tumor delineation (see also Figure 3), and also suggested that the gain of motion correction was more pronounced when PSF modeling is incorporated and vice versa. There have been studies in several oncologic applications reported^{15,38,39} that the respiratory compensation improved PET quantitative accuracy as compared with the no motion correction method in several aspects including contrast-to-noise ratio (CNR) (increased by 19–190%), the peak standardized uptake value (SUV) and maximum SUV (increased on average by 23.1% and 34.5%), the lesion size (reduced by 60.4% on average), and the lesion position (change of 60.9%). In comparison with the gating method, the motion correction method show improvement on the CNR by 6–51%.¹⁵

5. Summary

Motion degrades image quality and quantitation of PET images, and is an obstacle to quantitative PET imaging. Simultaneous PET-MR offers a tool which can be used for correcting the motion in PET images by using anatomical information from MR imaging acquired concurrently. Motion correction can be performed by transforming a set reconstructed PET images into the same frame or by incorporating the transformation into the system model and reconstructing the motion corrected image. Several phantom and patient studies in cardiac, oncologic phantoms demonstrated the significant improvement of PET quantitative accuracy such as CNR, SUV, and lesion detectability by motion correction as compared with the gating or the no motion correction methods. Therefore, MR-based motion correction has shown great promise to make quantitative PET imaging on simultaneous PET-MR possible.

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KEY POINTS

- Simultaneous PET-MR offers a tool which can be used for correcting the motion in PET images by using high quality anatomical information from MR imaging.
- The MR-based PET motion correction consists of two parts: estimating deformation field or motion field from MR data, and creating a motion-free PET images using the estimated deformation field.
- Several phantom animal and patient studies have validated that MR-based motion correction strategies have great promise for quantitative PET imaging in simultaneous PET-MR.

(a) Reconstruct-transform-average (RTA) method

(b) Motion-compensated image reconstruction (MCIR) method

Figure 1:

Diagram for (a) the Reconstruct-Transform-Average (RTA) and (b) MCIR methods for PET motion correction

Figure 2:

Slices of PET images from a moving cardiac phantom study comparing the images without motion correction and the motion corrected images (without and with PSF modeling).

(Adapted with permission from Petibon et al. Cardiac motion compensation and resolution modeling in simultaneous PET-MR: a cardiac lesion detection study. *Physics in Medicine and Biology*, 2013;58(7):2085–2102.)

Figure 3:

Slices of PET images from a patient study comparing the images without motion correction and the motion corrected images (without and with PSF modeling). Notice the heterogeneity of uptake in the lesion (indicated by the arrow in the T1-weighted MR image) revealed in motioned corrected images. (Adapted with permission from Petibon et al. Relative role of motion and PSF compensation in whole-body oncologic PET-MR imaging, *Med Phys* 2014; 41:042503.)