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Reduction of motion artifacts for PET imaging by respiratory correlated dynamic scanning

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Abstract

Organ motion caused by respiration is a major challenge in positron emission tomography (PET) imaging. This work proposes a technique to reduce smearing in PET imaging caused by respiratory motion. Dynamic scanning at 1 frame/s is used. A point source, used as a marker, is attached to the object's abdomen during the scan. The source position in the projection view moves with respiratory motion and can be used to represent the respiratory phase within the time interval in which each frame data are acquired. One hundred and twenty frames are obtained for each study. The range of the positions of the marker is divided into four groups, representing different respiratory phases. The frames in which the organ positions (phases) are the same summed to produce a static sub-sinogram. Each sub-sinogram then undergoes regular image reconstruction to yield a motion-free image. The technique is applied to one volunteer under both free and coached breathing conditions. A parameter called the volume reduction factor is adopted to evaluate the effectiveness of this motion-reduction technique. The preliminary results indicate that the proposed technique effectively reduces motion artifacts in the image. Coached breathing yields better results than free breathing condition. The advantages of this method are that (1) the scanning time remains the same; (2) free breathing is allowed during the acquisition of the image; and (3) no user intervention is required.

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Keywords: Respiration; PET; Motion artifact; Motion correction; Dynamic scan

1. Introduction

Patient movement due to respiration is known to influence lung, breast and abdomen treatments in radiation oncology. Respiration requires that the treatment volume is increased, such that a larger normal tissue volume is irradiated to higher doses. Osman et al. [1] reported that patient motion causes the fusion of the image sets to be imperfect. Osman et al. [2] investigated the respiratory motion artifacts in PET emission images obtained following CT attenuation correction, and concluded that respiratory motion artifacts were noted in images of most patients as a curvilinear cold area at the lung/diaphragm interface.

The time taken to perform a PET scanning significantly exceeds the time for which a patient can hold his or her breath, and respiratory motion artifacts importantly degrade image quality. Motion artifacts have two main effects [3–5]. First, it affects the accuracy of quantitation and reduces the measured standard uptake value (SUV). Second, it causes the lesion volume to be overestimated and increases the planned target volume and, therefore, the radiation dose delivered to the normal tissues. Eliminating the artifacts of respiration motion improve patient care.

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Kubo and Hill [6] examined several sensors including thermistors, thermocouples, strain gauges and pneumotachographs to determine the optimal sensor for monitoring respiratory motion. They used a respiratory gating system to monitor the motion of the chest wall by tracking the vertical position of two passive reflective markers which were fixed on the patient's abdomen. The motion was monitored and tracked using an infrared video camera. Beyer et al. [7] studied the effect of respiratory motion on the combined image quality in the PET/CT. They found that when limited breath holding was adopted, the frequency of severe artifacts in the area of the diaphragm was by halved, and the spatial extent of respiration-induced artifacts was by at least 40% lower than that obtained using acquisition protocols without any breathing instructions.

Nehmeh et al. [8] reported the use of a point source attached to the patient as a marker during PET scanning. The position of this source was used to track respiratory motion through the dynamic frames. The data were acquired in consecutive 1 s frames. All of the frames were reconstructed. A square region of interest (ROI) at a user-selected position was drawn for each image. The frames in which the point source fell within the ROI were identified. The sinograms that correspond to the same frames were added and then reconstructed. This approach has the following drawbacks: (1) the reconstruction for many 1 s frames is very time consuming, (2) user intervention is required to track the position of the point source, and (3) the reconstructed images are noisy and tracking the respiratory position is difficult.

This work proposes a method for reducing respiratory motion that is similar but better than that of Nehmeh et al. [8]. The position of the marker is obtained from the projection data rather than from the reconstructed image to overcome the disadvantages of their method.

2. Method

The relative position of the organs is assumed to be remained as the same when a small time bin is used. The frame data acquired within each time bin are regarded as motion free. The breathing cycle can be categorized into groups by the position of the organ determined from the projectional view data. The sinograms in each group are summed and reconstructed to yield an image with fewer motion artifacts.

A point source is attached to the abdomen of the subject during the scan. A dynamic 1-s-long scan is performed for consecutive 120 frames. The marker appears as a hot spot in the sinogram and its position changes with the respiratory phase. The position of the center of the marker in each sinogram is detected automatically from the contour of the hot spot. The breathing cycle can be separated into groups based on the marker position at each bin. The range of these locations is adopted to define the regions of M (= 4) groups, as shown in Fig. 1. The frames in which the marker positions fall in the same group are added to yield a new sub-sinogram. Since the time bin in each sub-sinogram is smaller than the period of the breathing cycle, the respiratory motion artifacts in the sub-sinograms are expected to be reduced from the uncorrected one. The outputs are M images, each corresponding to one position of the organ during respiratory motion. Notably, summing all of the M sub-sinograms yields the original uncorrected sinogram.

The projection view in the LA direction ($\theta = 90$), in which the respiratory motion is most obvious, is used to assign to the group. All projection data at other projection angles are processed based on the grouping of the LA view data.

Each sub-sinogram is reconstructed as a $128 \times 128 \times 15$ matrix (with a voxel size of $4.3\,\text{mm} \times 4.3\,\text{mm} \times 6.2\,\text{mm}$) using an order subset expectation maximization (OSEM)

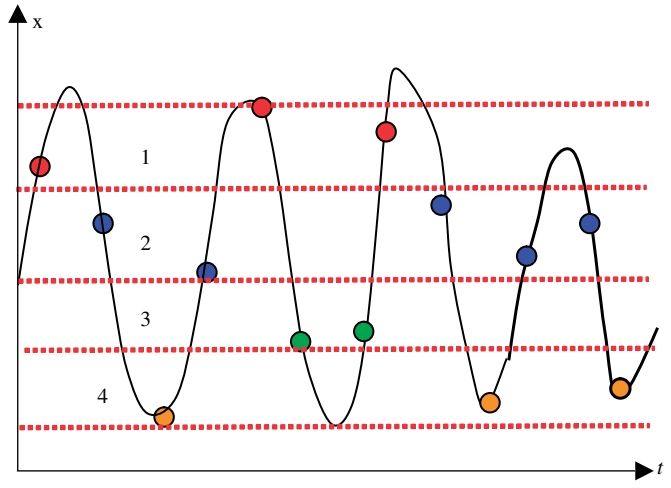

Fig. 1. The circles represent the marker position at the dynamic scan. The range of the marker position is divided into four groups.

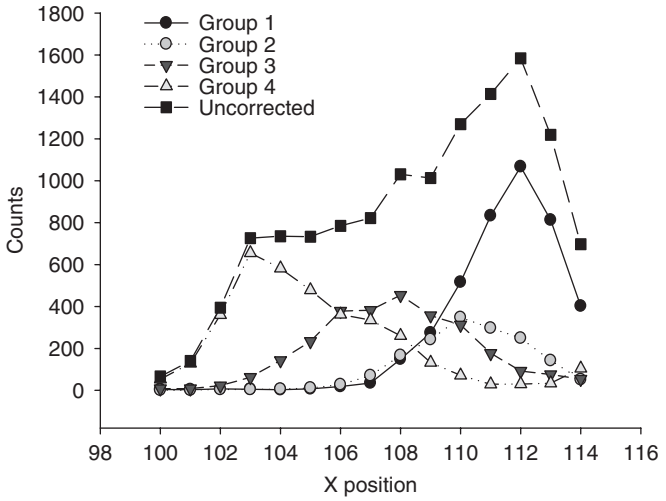

Fig. 2. Profiles drawn at the projection angle $\theta = 90$ for the projection data under coached breathing condition.

algorithm. The corrected-PET images were evaluated with reference to the non-corrected PET images using a factor called the volume reduction factor (VRF). The VRF is defined as the percentage difference between the apparent tumor volumes in the corrected and non-corrected images:

$$\mathrm{VRF} = \frac{V_t - \sum_{i=1}^{M} w_i V_i}{V_t} \quad (1)$$

where V_i is the volume and w_i (= number of frames in the ith group/120) is the weighted percentage in the ith group. The volume is defined as the sum of the areas in which the activity concentration in the lesion is under 42% of the maximum value. VRF detects the morphological change of the lesion due to the motion correction. A larger VRF value corresponds to larger differences caused by motion correction.

Informed consent was obtained from the volunteer. Nine glass tubes (with an inner diameter of 4 mm and a length of 1 cm), carrying point sources, were placed on the chest wall of the volunteer. The source had an approximate activity of 0.4 mCi/cm^3. The investigation was conducted using a CTI ECAT EXACT HR + scanner from Siemens. The PET data were acquired using consecutive 1 s frames in standard dynamic scanning mode with the scanner software. For

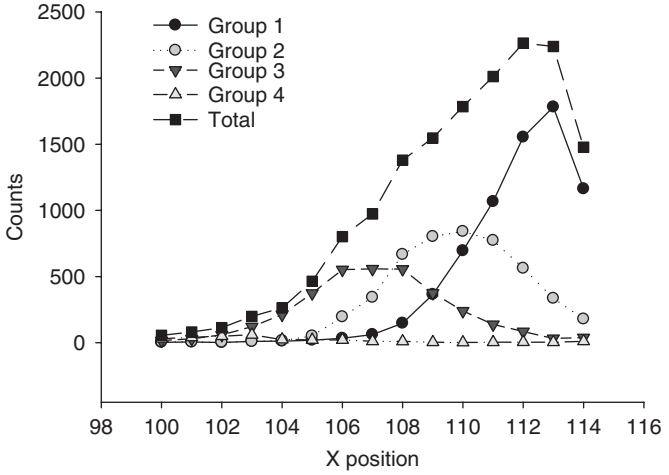

Fig. 3. Profiles drawn at $\theta = 90$ for the projection data of the volunteer under free breathing condition.

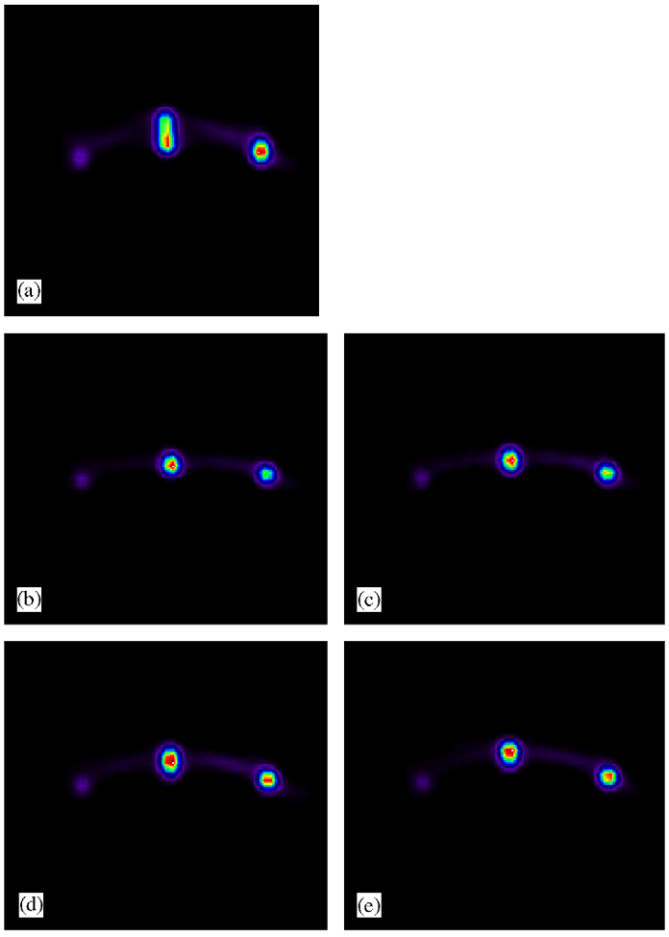

Fig. 4. (a)The uncorrected image and (b)–(e) the four groups of corrected images under coached breathing condition.

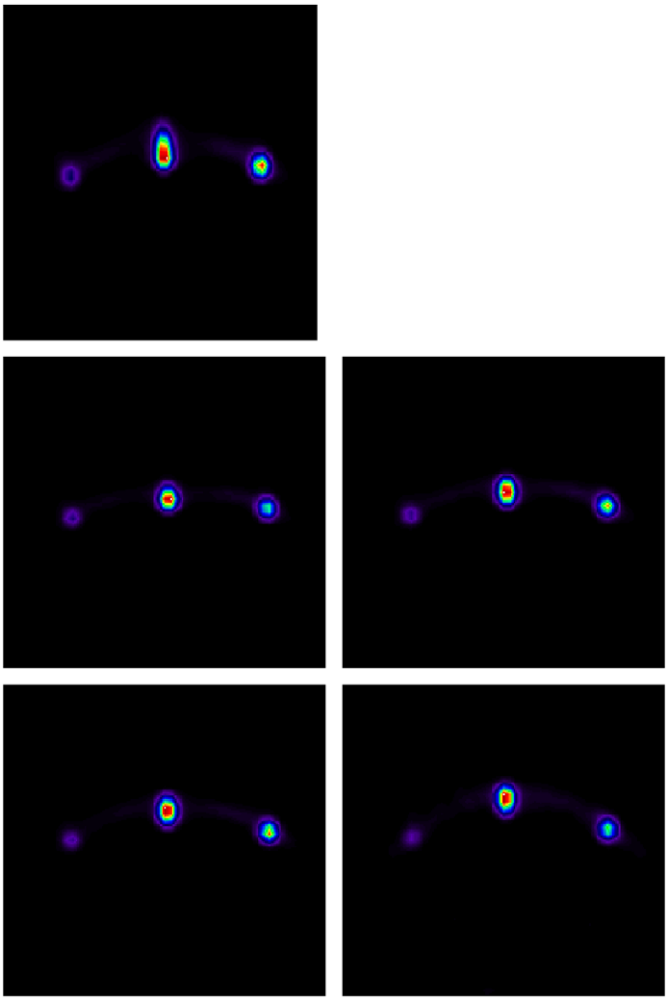

Fig. 5. (a)The uncorrected image and (b)–(e) the four groups of corrected images under free breathing condition.

Table 1
The VRF measured at the nine point sources for free breathing and coached breathing conditions

Source	1	2	3	4	5	6	7	8	9	Mean ± stdev
Free	6.47	9.0	7.24	0.	16.02	8.6	9.84	29.7	11.08	10.9 ± 7.8
Coached	3.81	12.86	−0.97	0.28	12.56	5.96	15.65	57.9	13.11	13.1 ± 16.7

demonstrative purposes, only 2-D scanning data (ring difference = 0) were stored to reduce the volume of the image.

Two breathing conditions were investigated in this study—free and coached breathing. A respiratory synchronization scheme was used to coach the breathing cycle. Synchronization was achieved when the volunteer adjusted his respiration on a visual cue. The inhalation and exhalation periods of the visual cues were both set to 2 s.

3. Results and discussions

Fig. 2 presents the profiles of the original (summation of 120 frames) sinogram and the four sub-sinograms for a point source at an angle $\theta = 90^{\circ}$, corresponding to right-to-left (lateral) projections for the breath-controlled section. The original profile (no correction) shows a significant blurring of the point source by respiratory motion. The proposed technique was used to separate the original profile into four Gaussian curves. Each curve represents the marker position at one phase in the respiratory cycle. The proposed method successfully separates the original data into motion projection data at various phases of the respiration. The size (= FWHM) of the point source was almost halved after motion correction. Most of the projection data are in the first and the fourth phases corresponding to the end of the exhalation and the end of the inhalation phases, respectively.

Fig. 3 presents a similar profile from the free-breathing data. Notably, the fourth group has very few data. This group contains mostly of the outliers caused by the irregular cycle in free breathing. Although the separations of the curve are not as large as under the coached breathing condition, the difference among groups remains visible. The separation can be improved by increasing the group number (M), but the number of frames in each region is then reduced and the noise in the reconstructed images increased.

Figs. 4 and 5 present the images reconstructed from the projection data under coached and free breathing conditions, respectively. Respiratory motion blurs the point sources in the uncorrected images (Figs. 4(a) and 5(a)). However, the proposed method restores the point source in each of the corrected images.

The VRFs at the nine point sources and their weighted average are determined from the reconstructed images. Table 1 presents the VRF measured under free breathing and coached breathing conditions. The variation of the measurements is large because not all parts of the body move to the same extent. The lesion volume smeared by the respiratory motion was reduced. The motion correction is better under the coached breathing condition than under the free breathing condition. A separate study compares traces of the lung/diaphragm interface (images obtained from fluoroscopy) and a reflective marker placed on the chest wall (using an infrared video camera). The motion of the diaphragm is around three times larger than that of the chest wall. Therefore, the amount of reduction in the motion artifacts in the lung exceeds that of the point sources demonstrated herein.

In this study, the projection data are acquired in 1 frame/s. For the HR + scanner, each frame occupies 5 Mbytes storage. The amount of data storage for dynamic scanning is proportional to the total period of study. Data storage might pose a problem. Since each frame has few events, the use of the list mode in image acquisition in dynamic studies is more efficient.

4. Conclusion

This work presented an approach for reducing respiratory motion using a radioactive source attached to the chest as a reference. The main advantages of this method are (1) the scanning time remains constant, (2) free breathing is allowed during image acquisition, and (3) no user intervention is required.

Acknowledgments

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