Joint correction of respiratory motion artifact and partial volume effect in lung/thoracic PET/CT imaging

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Purpose: Respiratory motion artifacts and partial volume effects (PVEs) are two degrading factors that affect the accuracy of image quantification in PET/CT imaging. In this article, the authors propose a joint motion and PVE correction approach (JMPC) to improve PET quantification by simultaneously correcting for respiratory motion artifacts and PVE in patients with lung/thoracic cancer. The objective of this article is to describe this approach and evaluate its performance using phantom and patient studies.

Methods: The proposed joint correction approach incorporates a model of motion blurring, PVE, and object size/shape. A motion blurring kernel (MBK) is then estimated from the deconvolution of the joint model, while the activity concentration (AC) of the tumor is estimated from the normalization of the derived MBK. To evaluate the performance of this approach, two phantom studies and eight patient studies were performed. In the phantom studies, two motion waveforms—a linear sinusoidal and a circular motion—were used to control the motion of a sphere, while in the patient studies, all participants were instructed to breathe regularly. For the phantom studies, the resultant MBK was compared to the true MBK by measuring a correlation coefficient between the two kernels. The measured sphere AC derived from the proposed method was compared to the true AC as well as the ACs in images exhibiting PVE only and images exhibiting both PVE and motion blurring. For the patient studies, the resultant MBK was compared to the motion extent derived from a 4D-CT study, while the measured tumor AC was compared to the AC in images exhibiting both PVE and motion blurring.

Results: For the phantom studies, the estimated MBK approximated the true MBK with an average correlation coefficient of 0.91. The tumor ACs following the joint correction technique were similar to the true AC with an average difference of 2%. Furthermore, the tumor ACs on the PVE only images and images with both motion blur and PVE effects were, on average, 75% and 47.5% (10%) of the true AC, respectively, for the linear (circular) motion phantom study. For the patient studies, the maximum and mean AC/SUV on the PET images following the joint correction are, on average, increased by 125.9% and 371.6%, respectively, when compared to the PET images with both PVE and motion. The motion extents measured from the derived MBK and 4D-CT exhibited an average difference of 1.9 mm.

Conclusions: The proposed joint correction approach can improve the accuracy of PET quantification by simultaneously compensating for the respiratory motion artifacts and PVE in lung/thoracic PET/CT imaging. © 2010 American Association of Physicists in Medicine.

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I. INTRODUCTION

Combined PET/CT imaging plays an important role in the staging and response to therapy of various types of cancers. The use of PET/CT imaging in lung/thoracic cancer, however, is limited by the patient's respiratory motion artifacts^{1,2} and partial volume effects (PVEs),³ both of which result in inaccurate PET image quantification. There are two sources

of error that can exist due to respiratory motion artifacts; these are motion blurring effects and mismatch between the PET and the CT images. Motion blurring in PET images is primarily caused by object motion during the long scan duration, which is necessary in PET imaging, while the mismatch between the PET and CT is primarily due to the discrepancy between the scan durations of the two imaging

modalities.⁴ PVE, on the other hand, is mainly caused by the finite spatial resolution of PET scanners.⁵ Due to this limited resolution, PET images underestimate the radiotracer uptake when the tumor size is less than two to three times the scanner's spatial resolution and also cause blurring of the edges of the object.⁵

Several approaches to correct for respiratory motion artifacts in PET/CT imaging have recently been introduced. One such approach, known as 4D PET/CT, acquires PET and CT data in multiple time frames according to the respiratory phase or amplitude captured in each frame.^{6,7} Since each frame captures only a small portion of the whole motion, these 4D PET/CT images can be regarded as motion-free images. However, a major problem of this approach is that each single frame only contains a small amount of counts and is therefore characterized by low signal-to-noise ratio (SNR). In this regard, various deformable registration techniques have been proposed to register these frames to one another in an effort to generate a motion-free PET image with improved SNR.^{6,7} Another approach that has been proposed to achieve the same objective is to incorporate the motion vectors between the different frames as part of the reconstruction algorithm to generate a final motion-free image.^{8,9} All of these techniques, however, suffer from high patient x-ray exposure due to the additional 4D CT scan as well as inaccuracies in the image registration process due to the low SNR of each PET/CT frame. To overcome these drawbacks, techniques maximizing the acquired data in a single phase/amplitude bin, such as repeated breath-hold (deep-inspiration breath-hold) or increasing scan duration in a single bin, have recently been proposed and implemented.¹⁰⁻¹² With such techniques, PET/CT data are acquired in a single phase/amplitude while maximizing the duty cycle of this specific gate and therefore results in reduced patient x-ray exposure and no requirement of deformable registration. In all of these approaches, however, respiratory monitoring devices are used to acquire the patient's breathing waveform. Furthermore, signal processing devices may be required in some of these techniques to generate triggers in order to indicate the onset of each frame or gate.

As with motion correction techniques, several methods to compensate for PVE have also been proposed in PET/CT imaging. These methods can be divided into two categories: Those applied at the regional level and those applied at the pixel level.⁵ The PVE correction methods applied at the regional level do not yield PVE-corrected images but rather PVE-corrected regional values [e.g., max or mean activity concentration (AC) that is constant across a region], which are suitable for the quantification of tumor uptake but not for visual analysis. In this regard, the resultant PET images using the corrected values do not have the usual texture/appearance that PET images usually have and hence should not be used for visual assessment. Examples of the most commonly used methods that utilize this approach are the recovery coefficient¹³ and geometric transfer matrix^{3,14} largely due to their relative ease of implementation. Both of these methods strongly rely on the technique used to measure the tumor size and shape and its corresponding accuracy, and

therefore require precise delineation of the tumor since different tumor sizes and shapes may result in drastically different correction factors. On the other hand, the PVE correction methods applied at the pixel level such as deconvolution¹⁵ and multiresolution methods¹⁶ do not focus on calculating PVE-corrected ACs but rather generate a PVE-corrected PET image for visual assessment. Such methods either greatly suffer from the high noise level, which was introduced during the deconvolution process, or require a precise registration between the high and the low resolution imaging modalities, which is usually difficult to realize. These limitations of using pixel-based techniques can eventually deteriorate the visual assessment of the PET images and the accuracy of the PVE correction approaches. Other approaches that are used to correct PVE include fitting methods,¹⁷ modeling PVE during reconstruction,¹⁸ and projection-space-based approaches.¹⁹ These techniques are applied either at the regional or pixel level to correct for the PVE with varying degrees of success and limitations. However, none of these approaches can simultaneously correct for PVE as well as respiratory motion artifacts as applied to lung/thoracic cancer.

Recently, a combined motion and PVE correction approach was proposed by Wiemker *et al.*²⁰ and later refined by Apostolova *et al.*²¹ to correct for both motion blurring and PVE effects in pulmonary nodules. In that approach, the combined effects of motion blur and PVE were considered as a Gaussian function, which when convolved with a contour representing the lesion size and shape, as obtained from a corresponding CT scan, will result in a maximum correlation with the observed PET image. In this regard, the approach of Wiemker *et al.* searches for the best Gaussian function that fits the observed PET data. Such an approach assumes that the motion blurring is a Gaussian function (which is not usually the case) and requires that a manual registration step is performed between the PET and the CT data to maximize the proposed correlation. More recently, Apostolova *et al.*²¹ refined this approach by optimizing this search while taking into consideration the nonstationary resolution nature of PET scanners along the transverse direction. Nonetheless, the use of a Gaussian function to represent motion blur has been shown to result in large variations from the actual motion and could lead to large quantification errors.²² To our knowledge, no other investigation has considered the simultaneous correction of respiratory motion and PVE.

In this paper, we propose a novel joint motion and PVE correction approach (JMPC) to improve the accuracy of PET image quantification by simultaneously correcting for respiratory motion artifacts and PVE. Rather than assuming a Gaussian motion, as was done by Wiemker *et al.* and Apostolova *et al.*,^{20,21} the proposed method does not make any assumptions about the motion blurring function but rather treat the motion blurring function as an unknown and estimate it as well as the average tumor AC through a deconvolution algorithm. Furthermore, this approach does not necessitate any image registration step, which is required by the approach of Wiemker *et al.*²⁰ and Apostolova *et al.*,²¹ while at the same time has the ability to jointly correct for PVE and

motion blurring in the lung/thoracic PET/CT imaging. The proposed approach is applied at the regional rather than the pixel level to generate the corrected AC, and in this regard, the resultant PET images with the corrected values do not have the usual texture/appearance that PET images usually have and hence should not be used for visual assessment. The objective of this paper is to describe how such a joint correction approach is formulated and evaluate its ability to improve the accuracy of PET image quantification using phantom and patient studies.

II. MATERIALS AND METHODS

II.A. The joint correction approach

In this section, we propose a mathematical framework to correct for both respiratory motion artifacts and PVE in lung/thoracic PET/CT imaging. In PET/CT imaging, PET images are usually much more blurred when compared to the corresponding CT images. This blurring effect, known as PVE, is primarily caused by the finite spatial resolution of the PET scanners, which is not only limited by the detector size/design (PET intrinsic resolution) but also affected by the reconstruction parameters. In this regard, the resultant PET image can be modeled as the convolution between the point spread function (PSF) of the PET system and the true PET image (object) that would be obtained if no PVE existed,

$$I_{\text{obs}} = I_{\text{true}} \otimes \text{PSF} + n, \quad (1)$$

where I_{obs} is the observed PET image, I_{true} is the true PET image (object), n is additive noise, and $\otimes$ represents the convolution process. The PSF represents the point spread function of the system that includes the effects of both the intrinsic resolution of the PET scanner and the reconstruction algorithm.

Under the influence of respiratory motion, lung/thoracic lesions will look even more blurred along the tumor motion direction. Since the patient's respiratory motion is usually nonrigid, a single motion blurring function is insufficient to represent the motion blurring effect of the whole body since different body parts can have different motions. However, in this study, we will assume that all pixels within a volume of interest (VOI) that contains the tumor and its motion extent are characterized by rigid motion since the tumor and VOI are usually relatively small. In this regard, the motion blurring effect for this lung lesion can be modeled as an additional convolution between the PVE-blurred PET image and the motion blurring kernel (MBK) defined within the VOI,

$$\begin{aligned} I_{\text{obs}} &= (I_{\text{true}} \otimes \text{PSF} + n) \otimes \text{MBK} \\ &= I_{\text{true}} \otimes \text{PSF} \otimes \text{MBK} + n', \\ \text{where } n' &= n \otimes \text{MBK}. \end{aligned} \quad (2)$$

Equation (2) from here onward is referred to as the overall-blurring equation since it represents the overall-blurring effect in lung/thoracic PET imaging (PVE+motion).

For lung/thoracic lesions, the following assumptions are made: (I) The lesion size/boundary on the PET image corresponds to that of the CT image when not affected by PVE

and motion blurring, and (II) the background of the lesion is relatively uniform within the delineated VOI. The first assumption is routinely used to delineate lung nodules on PE images,²⁰⁻²³ while the effect of background variation can be minimized by reducing the size of the VOI around the tumor. Based on these assumptions, the true lung lesion image or object (I_{true}) can be written as

$$\begin{aligned} I_{\text{true}} &= (\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{IF} + \text{AC}_{\text{bkg}} \times U \\ &= (\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{IF} + I_{\text{bkg}}, \\ \text{where } I_{\text{bkg}} &= \text{AC}_{\text{bkg}} \times U, \end{aligned} \quad (3)$$

where IF represents the indicator function of the lung lesion that has unit values inside the lesion but zeros outside the tumor. The IF can be derived from the delineation of the tumor in the corresponding CT image. In addition, I_{bkg} , AC_{bkg} , and U represent the uniform background image, average background AC, and a unity image (all pixels equal to one), respectively. Furthermore, AC_{mean} represents the mean AC distribution of the lung lesion. When Eqs. (2) and (3) are combined with one another, we have

$$\begin{aligned} I_{\text{obs}} &= [(\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{IF} + I_{\text{bkg}}] \otimes \text{PSF} \otimes \text{MBK} + n' \\ &= (\text{IF} \otimes \text{PSF}) \otimes [(\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{MBK}] \\ &\quad + I_{\text{bkg}} \otimes \text{PSF} \otimes \text{MBK} + n' \\ &= (\text{IF} \otimes \text{PSF}) \otimes [(\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{MBK}] \\ &\quad + I_{\text{bkg}} + n', \quad \text{where } I_{\text{bkg}} \otimes \text{PSF} \otimes \text{MBK} \\ &\equiv I_{\text{bkg}} \end{aligned} \quad (4a)$$

(this equivalence is due to the uniformity of the background) or

$$\begin{aligned} (I_{\text{obs}} - \text{AC}_{\text{bkg}} \times U) &= (\text{IF} \otimes \text{PSF}) \\ &\quad \otimes [(\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{MBK}] + n'. \end{aligned} \quad (4b)$$

This equation shows that the observed PET image, following the subtraction of the background AC, which can be directly measured on the observed PET image, depends on the convolution between two terms. The first term, $\text{IF} \otimes \text{PSF}$, can be derived from the tumor delineation on the CT image and measuring the PET scanner resolution, respectively. The second term $((\text{AC}_{\text{mean}} - \text{AC}_{\text{bkg}}) \times \text{MBK})$, on the other hand, can be calculated by deconvolving the left-hand side ($I_{\text{obs}} - \text{AC}_{\text{bkg}} \times U$) by the first term on the right-hand side ($\text{IF} \otimes \text{PSF}$) of Eq. (4). In order to suppress the noise amplification, which is a characteristic of the deconvolution operation, we employ an expectation maximization (EM) deconvolution algorithm, which is an iterative approach;^{24,25} if

$$g(x, y, z) = h(x, y, z) \otimes f(x, y, z) + n(x, y, z),$$

then

FIG. 1. The workflow of the proposed joint correction approach.

$$f_{k+1}(x, y, z) = f_k(x, y, z) \times \left[h(-x, -y, -z) \otimes \frac{g(x, y, z)}{h(x, y, z) \otimes f_k(x, y, z)} \right], \quad (5)$$

where $f_k(x, y, z)$ is the estimation of $f(x, y, z)$ during the k th iteration and any multiplication and division are point-by-point within the image. In this paper, 20 iterations were selected based on previous publications²² and using computer simulations that showed this number to be sufficient to reach convergence (data not shown). In this deconvolution algorithm, a non-negativity constraint is first applied to guarantee that the term $(I_{\text{obs}} - AC_{\text{bkg}} \times U)$ is greater or equal to zero everywhere within the VOI. In the above equation, we had the following substitutes:

$$g(x, y, z) = I_{\text{obs}} - AC_{\text{bkg}} \times U, \quad (6a)$$

$$h(x, y, z) = \text{IF} \otimes \text{PSF}, \quad (6b)$$

$$f(x, y, z) = (AC_{\text{mean}} - AC_{\text{bkg}}) \times \text{MBK}. \quad (6c)$$

Since the MBK represents the probability of a tumor to appear in a specific location, its integration over the whole space/image/VOI should be equal to unity,

$$\int \int \text{MBK}(x, y, z) dx dy dz = 1. \quad (7)$$

In this regard, the mean activity concentration of the tumor AC_{mean} can be derived from the normalization of $f(x, y, z)$,

$$\begin{aligned} AC_{\text{mean}} - AC_{\text{bkg}} &= \int \int f(x, y, z) dx dy dz \\ &= \int \int [(AC_{\text{mean}} - AC_{\text{bkg}}) \times \text{MBK}] dx dy dz \end{aligned} \quad (8a)$$

or

$$AC_{\text{mean}} = AC_{\text{bkg}} + \int \int [(AC_{\text{mean}} - AC_{\text{bkg}}) \times \text{MBK}] dx dy dz, \quad (8b)$$

and the MBK can then be simultaneously calculated from

$$\text{MBK} = f(x, y, z) / (AC_{\text{mean}} - AC_{\text{bkg}}). \quad (9)$$

The whole process of the proposed joint correction approach is summarized in Fig. 1. In this approach, the tumor AC that is corrected for both respiratory motion and partial volume effect is estimated from a single PET/CT scan without the need for tumor motion monitoring devices for 4D-CT scans. The by-product of this approach, MBK, can be used for other applications such as motion control, motion extent estimation, and motion trajectory monitoring. In this paper, both phantom and patient studies will be performed to evaluate the ability of this approach to improve PET image quantification.

II.B. PET spatial resolution

A GE Discovery STE PET/CT scanner (GE Healthcare, Waukesha, WI) was used for the phantom study in this paper. The PET gantry of this scanner consists of 24 rings of 560 detector crystals per ring and the ring diameter is 88.6 cm. The PET component has a transaxial FOV of 70 cm and a 15.7 cm axial extent.

In order to obtain the spatial resolution of a PET image reconstructed using a specific set of parameters, a point source is usually imaged and its FWHM is measured after image reconstruction using a predetermined reconstruction parameter set. In this paper, however, a more practical and automated approach is employed to measure the PET resolution, which utilizes the routinely available uniform cylindrical phantom.²⁶ This resolution measurement approach enables the derivation of a scanner PSF from deconvolving the PET image from a binary image obtained from the corresponding CT image in Fourier transform space. The FWHM

FIG. 2. Spatial resolution measurement for the PET scanner. (a) The binary image obtained from the CT image, the PET image, and the derived PSF. (b) The radial and tangential line profiles across the PSF image and the Gaussian fitting.

of the derived PSF is then measured using a Gaussian fit to represent the PET image resolution. This approach has been tested to be robust and accurate.²¹ Figure 2 shows the results from the cylindrical phantom measurement for the scanner used in this paper. The binary image obtained from the CT image, the PET image, and the derived PSF image are shown in Fig. 2(a), while the results from the Gaussian fitting are shown in Fig. 2(b) for both the radial and the tangential line profiles through the PSF image. The results were based on the following reconstruction parameters: 3D OSEM (2 iterations, 20 subsets), 256 mm FOV, 256×256 matrix size, and 2 mm filtering. These parameters were chosen since our phantom images were generated using these parameters. The two Gaussian fittings indicate that the radial and tangential FWHM of the PSF are 4.5 and 4.7 mm, respectively. Therefore, in the phantom and patient studies of this paper, a PSF with 4.6 mm FWHM is used to represent the PET spatial resolution.

II.C. Phantom studies

The objective of the phantom studies is to evaluate the performance of the proposed joint correction approach in a controlled pseudoclinical environment. Two phantom studies each consisting of two moving spheres (33 and 22 mm in diameter for study 1, 13 and 10 mm in diameter for study 2) were used. In the first study, a linear sinusoidal waveform (2 cm peak-to-valley amplitude, 5 s period) was used to translate the two spheres in the left-right direction in the transaxial plane. In the second study, a 2D circular waveform (2 cm in diameter, 5 s period) was applied to translate the two spheres in a circular trajectory in the transaxial plane. The

linear motion waveform represented a regular lung lesion motion, while the circular waveform was used to simulate hysteresis. In the two phantom studies, the two spheres were filled with ^{18}F -FDG water with ACs of 39.7 and 16.2 kBq/cm³, respectively. No background AC, however, was included in the two phantom studies primarily due to a difficulty in the experimental setup whereby an actuator can be programmed to move the spheres in the desirable motion trajectory (particularly circular) within the PET transverse FOV, while the spheres are immersed in a water bath with the background AC. The effects of the background AC, however, are investigated in the patient studies (see Sec. II D). In this regard, the phantom studies can be considered as experiments that are conducted as a proof of principle for the proposed approach. The setup of the two phantom studies is shown in Figs. 3(a) and 3(b), respectively. In the top view of Fig. 3(a) the 22 mm sphere was hidden behind the 33 mm sphere. In the setup of the second phantom study, the two sinusoidal waveforms (horizontal and vertical) had a 90° phase shift in order to ensure that the spheres traversed a circular motion trajectory.

In both studies, the spheres were placed in the central FOV of the PET/CT scanner and a CT scan was first captured when the two spheres were stationary. PET data were then acquired using 3D, mode while the two spheres moved according to the linear and circular waveforms. For the linear sinusoidal motion only, an additional static PET scan was acquired, while the two spheres remained stationary during the data acquisition. The aim of this static PET scan was to generate a PET image without any motion blur (PET image with PVE only) for reference purposes. The duration of each

Setup of Phantom Studies

FIG. 3. The setup of the phantom studies: (a) Linear sinusoidal motion and (b) circular motion for both side and top views.

PET scan was set to 3 min based on the standard clinical protocol at our institution. After data acquisition, the PET and CT images were reconstructed using OSEM (2 iterations, 20 subsets, and 2 cm filter) and FBP, respectively. The matrix size used for PET and CT reconstructions were 256×256 and 512×512 , respectively. The PET image was then upsampled to a 512×512 size corresponding to the CT image. All reconstructions were performed using a FOV of 256 mm.

For each sphere in each study, a cubic VOI was drawn around the sphere and an IF was delineated within the defined VOI based on the CT image. The sphere segmentation was based on CT numbers of ± 8 that represented water, which was then further refined by the comparison between the segmented and the true sphere volumes. The derived IF as well as the measured scanner PSF in Sec. II B were then incorporated into Eqs. (4) and (5) to estimate the average tumor AC and MBK. In both phantom studies, the background AC was set to zero since no AC was placed in the background. For reference, the maximum and mean AC of both spheres on the PET image with PVE only (linear sinusoidal motion only) and the PET image with both motion blurring (linear and circular motion) and PVE were also calculated. Furthermore, the derived MBK was compared to the true MBK and a correlation coefficient was determined. The motion extent of the spheres was also derived from the MBK using a threshold which results in 80% of the overall probability in the MBK image. This threshold has been tested to be optimum using simulations as it can generate the same motion extent as the true motion.

II.D. Patient studies

The objective of the patient studies is to evaluate the performance of the proposed joint correction approach in a real clinical environment where background AC is present and corresponds to the mathematical framework in Sec. II A. An Institutional Review Board approval was first acquired prior to the patient studies.

Eight patients (four male and four female; age: 56 ± 8 yr old) referred for PET/CT and 4D-CT evaluation of nonsmall cell lung cancer (NSCLC) lesions were selected to test the performance of the proposed joint motion-PVE correction approach. All patients fasted for 4 h prior to being injected intravenously with 296–444 MBq (8–12 mCi) of ^{18}F -FDG. Imaging started 60–90 min postinjection. The whole imaging process consisted of a regular whole-body CT scan, a whole-body PET scan, and a CINE CT scan of one bed position which covered the patient's NSCLC lesion. The duration of the PET scan was selected to be 3 min per bed position, while the CINE CT scan lasted one respiratory cycle plus 1 s. During the whole process, the patients were requested to breathe regularly.

The PET data for each patient were reconstructed using a 3D OSEM algorithm (2 iterations, 20 subsets) after attenuation correction by the CT image. A rectangular VOI was first drawn to enclose the NSCLC lesion on the PET image, while the same VOI was copied to the CT image. An IF was then delineated on the CT image within the rectangular VOI to represent the position and size of the lesion using a threshold of -425 HU, which was facilitated by manual delineation. The background AC was then calculated based on the aver-

Phantom Study 1: Linear Sinusoidal Motion

FIG. 4. Results from phantom study 1. The two spheres (top row: 33 mm; bottom row: 22 mm) on (a) the PET image with motion blurring and PVE, (b) the PET image with PVE only, and (c) the PET image generated using the joint correction method. The estimated MBK from the joint correction approach is shown in (d).

age pixel values of 50 randomly selected pixels within the background of the PET rectangular VOI. The IF, the background AC, as well as the scanner PSF were then incorporated into the overall-blurring equation (4) to estimate the mean tumor AC and MBK. The percent increases in mean/max AC after joint correction were then calculated based on dividing the difference between the estimated AC and the mean/max AC in the uncorrected image by the mean/max AC in the uncorrected image, respectively. The motion extent of the spheres was also derived from the MBK using a threshold which results in 80% of the overall probability in the MBK image. Furthermore, the acquired CINE CT data were processed using an in-house software to generate ten phases of 4D-CT data. A second motion blurring function was then derived by monitoring the displacement of the tumor centroid among the ten different phases. The extent of this motion blurring function was then compared to the motion extent of the MBK derived from the proposed joint correction approach.

III. RESULTS

III.A. Phantom studies

The PET images with both motion blurring and PVE, PVE only, and the corrected image using the proposed joint correction method for the linear sinusoidal phantom study are shown in Figs. 4(a)–4(c), respectively, for both spheres (top row: 33 mm spheres; bottom row: 22 mm spheres). The PET images with both motion blurring and PVE, and the

corrected image using the proposed joint correction method for the circular phantom study, on the other hand, are shown in Figs. 5(a) and 5(b), respectively, for both spheres (top row: 13 mm spheres; bottom row: 10 mm spheres). For each study, all the PET images are normalized to the same color scale for display. Note that the images in Figs. 4(c) and 5(b) do not represent a usual PET image but rather a corrected AC value multiplied by an IF, which is used for visual display only. Quantitative results of the AC for both studies in all the PET images are shown in Table I [(a) and (b), respectively]. This table indicates that the average AC derived from the joint correction approach approximates the true AC very well with an average difference of only 1% and 3% for the two phantom studies, respectively. The other two PET images (with PVE only and with both motion blurring and PVE), however, exhibit an average of 25% and 52.5% (90%) decrease in their average AC from the true values for the linear sinusoidal (circular) motion phantom study, respectively. Note that no results are presented for the image with PVE only in the circular motion case since no PET data were acquired under this condition.

The estimated MBK from the proposed joint correction approach is displayed in Figs. 4(d) and 5(c) for the two studies, respectively. The MBK in Fig. 4(d) has a 2 cm amplitude and also shows a nonuniform distribution consistent with the sinusoidal motion profile that was applied to the spheres. The correlation coefficient between the estimated and the true MBK is 0.92, indicating that the estimated MBK approximates the true MBK very well. On the other hand, the MBK

Phantom Study 2: Circular Motion

FIG. 5. Results from phantom study 2. The two spheres (top row: 13 mm; bottom row: 10 mm) on (a) the PET image with motion blurring and PVE, and (b) the PET image generated using the joint correction method. The estimated MBK from the joint correction approach is shown in (c).

in Fig. 5(c) has a 2 cm amplitude circular function, consistent with the circular motion profile that was applied to the spheres. The correlation coefficient between the estimated and true MBK in this study is 0.89.

III.B. Patient studies

The coronal view of the CT image, PET image, and the derived MBK after applying the proposed joint correction technique are shown in Figs. 6(a)–6(c), respectively, for one of the patient studies. A VOI (black circle) is drawn on the

CT image to delineate the tumor boundary, which was used to generate the IF. A comparison of the tumor shape and location between the CT and PET images [Figs. 6(a) and 6(b)] shows that the tumor is elongated along the cranial-caudal direction (Z-axis) on the PET image. This result can be verified by the MBK displayed in Fig. 6(c), which exhibits the largest motion along the cranial-caudal direction (Z-axis) with a motion extent of 11 mm. The maximum and mean SUVs of this tumor on the PET image before the joint correction were 6.17 and 1.73, respectively, while the SUV (mean) after applying the joint correction technique increases to 6.52, indicating that the proposed joint correction approach has the potential to increase the accuracy of the PET quantification. Only mean SUV is reported following the application of the joint correction technique since this approach only generates corrected mean values.

The tumor SUVs of all the eight patient studies are summarized in Table II(a) before and after applying the proposed joint correction technique. The increase in maximum/mean SUV for the eight patient studies are also summarized in Table II(b). These tables clearly show that the maximum and mean SUV are increased by an average of $125.9\% \pm 299\%$ and $371.6\% \pm 432\%$, respectively. Among the eight patient studies, patient 3 has the largest increase in SUV primarily due to its relatively small tumor size and large motion extent. The motion extents measured from both the joint correction technique and 4D-CT are also shown in Table II(a) and their differences are listed in Table II(b). The two tables show that the two measured motion extents have an average difference of 1.9 ± 1.7 mm, indicating that the proposed joint correction technique can have similar results when measuring the tumor motion extents as the 4D-CT technique.

IV. DISCUSSION

In this paper, we proposed a novel approach to improve the accuracy of PET image quantification by jointly correcting for respiratory motion artifacts and PVE in patients with lung/thoracic cancer. This method incorporates a model for PVE, motion blurring, and object size/shape to estimate the true tumor AC and MBK using an EM deconvolution algorithm. In order to test the performance of this approach, two

TABLE I. (a) Results from phantom study 1: Linear sinusoidal motion. (b) Results from phantom study 2: Circular motion (data for PET image with PVE only do not exist in this study). (All data are normalized to the true AC.)

	True	Joint correction	PET image with PVE only	PET image with motion and PVE
(a) Results from phantom study 1				
Average AC (33 mm sphere)	1.00	0.99	0.77	0.52
Average AC (22 mm sphere)	1.00	0.99	0.73	0.43
Maximum AC (33 mm sphere)	1.00	N/A	0.95	0.89
Maximum AC (22 mm sphere)	1.00	N/A	0.85	0.72
(b) Results from phantom study 2				
Average AC (13 mm sphere)	1.00	0.97		0.11
Average AC (10 mm sphere)	1.00	0.97		0.09
Maximum AC (13 mm sphere)	1.00	N/A		0.20
Maximum AC (10 mm sphere)	1.00	N/A		0.14

FIG. 6. Results from one patient study: (a) The coronal view of the CT image, (b) the coronal view of the PET image, and (c) the derived MBK.

phantom studies and eight patient studies were conducted. The results from these studies showed that the corrected AC and the derived MBKs approximated the true injected AC and the true motions very well.

The proposed joint correction approach is classified among those techniques that are applied at the regional rather than the pixel level.⁵ In this regard, this approach produces a corrected AC or SUV *value* rather than a corrected PET *image*. In other words, the proposed joint correction approach should be utilized to improve the accuracy of PET lesion quantification (value) and not to generate a corrected image that is suitable for a physician's evaluation/interpretation. This is primarily due to the fact that the resultant images do not exhibit the usual texture/appearance normally seen in PET images since all voxel values within a region have the same value.

One of the main advantages of this proposed technique is that it can correct for respiratory motion artifacts and PVE without the need to acquire the patient's respiratory waveform and therefore does not necessitate the use of motion monitoring devices. Furthermore, unlike other motion correction techniques, this approach does not require a 4D-CT scan and therefore does not result in high patient x-ray exposure nor require the use of deformable registration. Another advantage of this technique is that the trajectory and extent of the lesion motion (which are important components for radiation therapy planning using PET/CT) can be deter-

mined from the MBK, which is the by-product of the proposed approach, since it represents the probability distribution of the tumor position at a specific location. The caveat in this advantage, however, is that the direction of motion cannot be determined from the MBK.

Wiemker *et al.*²⁰ and Apostolova *et al.*²¹ proposed a similar approach to correct for respiratory motion blurring and PVE in PET/CT imaging, which also neither requires the acquisition of the patient's motion waveform nor a 4D-CT scan. Their approach, however, necessitates a local registration refinement step between the PET and CT images to improve the accuracy of the resultant motion and PVE-corrected PET image, which, in turn, could be affected by the noise content in the PET image during the registration process. Our proposed approach, on the other hand, does not require such a registration step but rather automatically resolves any such mismatch during the joint correction process, which is represented by the probability distribution of the derived MBK. Figure 4 proves this capability by showing that although the PET and CT were acquired with a 1 cm mismatch [Fig. 4(d) shows that the average location of the MBK has a 1 cm position shift with respect to the CT], the results of the proposed joint motion and PVE correction technique can automatically resolve this mismatch accurately, as shown in Fig. 4(c) and Table I(a). Furthermore, the approaches of Wiemker *et al.*²⁰ and Apostolova *et al.*²¹ assume that the motion blurring in the PET image can be ap-

TABLE II. (a) SUV and motion extents of the patient studies. (b) Comparison.

(a) AC/SUV and motion extents						
Patient ID	Tumor size (cm ³)	SUV after joint correction	Before joint correction		Motion extent (mm)	
			Max SUV	Mean SUV	Joint correction	4D-CT
1	21.1	21.3	15.6	8.8	11	10.8
2	15.3	12.9	18.2	8.2	8	7.8
3	1.1	20.9	2.18	1.43	15	10.3
4	1.0	6.52	6.17	1.73	11	6.6
5	1.6	5.99	6.43	2.61	7	6.3
6	1.5	13.9	10.4	2.3	5	2.4
7	0.7	16.6	7.44	3.15	5	4.6
8	10.3	11.5	13.5	6.6	7	8.6

(b) Comparison			
Patient ID	Increase in maximum SUV (%)	Increase in mean SUV (%)	Difference of tumor volume (mm) (joint correction vs 4D-CT)
1	36.5	142.0	0.2
2	-29.1	57.3	0.2
3	858.7	1361.5	4.7
4	5.7	276.9	4.4
5	-6.8	129.5	0.7
6	33.7	504.3	2.6
7	123.1	427.0	0.4
8	-14.8	74.2	1.6
Average	125.9 ± 299	371.6 ± 432	1.9 ± 1.7

proximated by a Gaussian blurring function. An analysis of various patient's respiratory motion, however, shows that the MBK or image blurring due to motion cannot be described using a Gaussian function but rather using a function that is caused by a near-sinusoidal respiratory motion.²⁷ Furthermore, since different patients have different respiratory styles and can exhibit different motion blurring effects, the MBK cannot be generalized by restricting its shape. Our joint correction approach, on the other hand, does not impose any specific restrictions to represent the MBK but rather treats this MBK as an unknown in the overall-blurring equation and estimates it during the joint correction process. A comparison (data not shown) between the proposed joint correction approach and the approaches of Wiemker *et al.* and Apostolova *et al.* using our phantom data shows that the joint correction approach performs better at estimating the tumor AC (average of $2.0 \pm 1.0\%$ difference from true values) versus the approach of Wiemker *et al.* (average: $5.5 \pm 0.5\%$ and $42 \pm 27\%$ difference from true values for linear motion and circular motion, respectively) and has the added capability of estimating the motion blurring function.

To implement the proposed joint correction technique, we made the assumption that the lesion size/boundary on the CT image matches that on the corresponding PET image when not affected by PVE and motion blurring. This assumption is routinely used for characterizing lung nodules^{20–23} and allowed us to accurately delineate the tumor boundaries on the CT rather than on the PET image to derive the IF necessary to calculate the corrected mean AC and MBK. The delineation

of the tumor boundary from the CT image was relatively easy to accomplish and was automated in the phantom studies since the CT numbers of the spheres were largely different from those of the background. In the patient studies, however, this process could be more difficult to achieve, especially when the lesion is attached to other organs, such as the lung wall, hilum, or diaphragm. In such cases, manual delineation was used to aid in this process.

The results from the phantom and patient studies indicate that the proposed joint correction approach has the potential to increase the accuracy of PET quantification. This improvement in quantification accuracy can be verified in the phantom studies since the estimated sphere AC after applying the joint correction technique approximates the true sphere AC very well. In patient studies, however, the lack of true tumor AC/SUV prevents the precise evaluation of the joint correction technique. However, since the tumor SUV was increased after applying the joint correction approach, we believe that this technique has the potential to improve the accuracy of PET quantification in clinical environment as both PVE and respiratory motion have an effect of decreasing the tumor SUV. Furthermore, although the motion extents derived from the joint correction approach and 4D-CT had an average difference of 1.9 mm only, some patients had a difference of >4 mm. This large discrepancy is primarily due to the differences in the number of acquired respiratory cycles during the PET and 4D-CT scans. PET scans are acquired over multiple respiratory cycles; in this regard, the measured motion extent from the joint correction approach would represent the

averaged motion extent within these cycles, while the motion extent during 4D-CT is determined from a single respiratory cycle and hence can be greatly different from that determined from PET. In this regard, a discrepancy between the results from the proposed joint correction approach and the 4D-CT acquisition is expected. The 4D-CT results in this paper can only be used for reference or comparison purposes rather than “gold standard.” In addition, we believe that the motion function derived from the PET image by using the joint correction approach is a better representation of the patient’s motion when compared to the motion function derived from 4D-CT since PET captures the averaged motion function of the patient while 4D-CT only captures one respiratory cycle.

One limitation of the proposed approach is that the PSF used in the overall-blurring equation is derived from a specific scanner and can only be used for a specific combination of reconstruction parameters: 3D OSEM, 2 iterations, 20 subsets, 256 mm FOV, 256×256 matrix size, and 2 mm filtering. For other scanner models and reconstruction parameters, a different PSF has to be derived. A practical method to do this is to create PSF look-up tables corresponding to various combinations of scanner models and reconstruction parameters. PSFs for reconstruction parameters that are not part of a look-up table can then be interpolated or extrapolated from the existing data points. Furthermore, the resolution measurement technique²⁶ used in this paper only measures the scanner resolution at a specific location corresponding to the radius of the cylindrical phantom (10 cm off center) and hence does not accurately represent the nonstationary nature of the transverse resolution of PET scanners. In this regard, the accuracy of tumor quantification following joint correction will be affected in lesions that are located at radial distances other than 10 cm. However, since the change in transverse resolution across the FOV of current PET scanners is less than 1 mm FWHM,²⁸ this discrepancy is expected to have very small effects. However, in order to completely remove the effects of scanner resolution variation, different PSF with different FWHM could be used, depending on the location of the tumor during the joint correction process.

Another limitation of the proposed approach is its implicit assumption that the tumor is confined to rigid motion only throughout the CT and PET data acquisitions. This assumption is necessary to ensure that the contour corresponding to the lesion segmentation from the CT image maintains its same shape and size on the PET image. Such an assumption is valid for solid lesions that do not undergo any deformations during the patient’s respiratory cycle. For nonsolid lesions, however, the tumor AC will be over- or underestimated, depending on the size and shape of the tumor captured during the CT image acquisition. This over- or underestimation in tumor AC eventually can lead to inaccurate PET image quantification. One approach to solve this problem is to acquire phase matched CT and PET images since such a process can capture a similar tumor size and shape during the two imaging modalities. This approach, however, suffers from the same problems as 4D PET/CT acquisition and therefore is not recommended in this paper.

V. CONCLUSION

We proposed a joint correction approach that can simultaneously compensate for the respiratory motion artifacts and PVE in lung/thoracic PET/CT imaging. It has the potential to increase the accuracy of PET quantification without the need to measure the motion blurring function during data acquisition. This approach can only generate a motion and PVE-corrected tumor AC rather than a motion-PVE-corrected PET image and therefore is classified among those techniques that are applied at the regional rather than the pixel level.

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