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Limits of Ultra-Low Dose CT Attenuation Correction for PET/CT

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Abstract

We present an analysis of the effects of ultra-low dose X-ray computerized tomography (CT) based attenuation correction for positron emission tomography (PET). By ultra low dose we mean less than approximately 5 mAs or 0.5 mSv total effective whole body dose. The motivation is the increased interest in using respiratory motion information acquired during the CT scan for both phase-matched CT-based attenuation correction and for motion estimation. Since longer duration CT scans are desired, radiation dose to the patient can be a limiting factor. In this study we evaluate the impact of reducing photon flux rates in the CT data on the reconstructed PET image by using the CATSIM simulation tool for the CT component and the ASIM simulation tool for the PET component. The CT simulation includes effects of the x-ray tube spectra, beam conditioning, bowtie filter, detector noise, and beam hardening correction. The PET simulation includes the effect of attenuation and photon counting. Noise and bias in the PET image were evaluated from multiple realizations of test objects. We show that techniques can be used to significantly reduce the mAs needed for CT based attenuation correction if the CT is not used for diagnostic purposes. The limiting factor, however, is not the noise in the CT image but rather the bias introduced by CT sinogram elements with no detected flux. These results constrain the methods that can be used to lower CT dose in a manner suitable for attenuation correction of PET data. We conclude that ultra-low-dose CT for attenuation correction of PET data is feasible with current PET/CT scanners.

I. Introduction

In some applications of PET/CT imaging, there is increased interest in using respiratory motion information acquired during the CT scan for both phase-matched CT-based attenuation correction of the PET data and for motion estimation during the respiratory scan [2,9]. Since the respiratory cycle is typically 4-6s and the motion range is roughly the apex of the lung to the lower abdomen, the radiation dose delivered by even a ‘low-dose’ CT scan (typically 50-100 mAs) can be a limiting factor for diagnostic imaging [1,8]. We are investigating methods to reduce the photon flux in the CT scan while still providing CT images adequate for attenuation correction. As part of that effort we are evaluating the impact of the lower limits of X-ray photon flux suitable for PET attenuation correction.

The CT images are used for attenuation correction of the PET emission data [16] as shown in Figure 1. As a result, bias and noise effects in the CT images, characteristic of low dose scans, can be expected to propagate through to the PET image in a non-linear fashion.

There are several potential methods for reducing dose in CT scans as listed in Table 1. In this effort we are assuming that the properties of X-ray tubes in clinical scanners are not easily modified, so that other aspects of the CT sub-system must be modified, or at least in operation on the scanner.

As a necessary counterpart, there are several potential methods to reduce the bias and noise, which are characteristic of low dose CT images, as summarized in Table 2. Potentially the

most important of these is the recognition that the requirements for CT-based attenuation correction images are dramatically less than those for diagnostic CT images [12,16].

The goal of our study is to gain an understanding of the impact of reduction of CT dose on the bias and noise in a PET image. To accomplish this we use a series of simulation studies as described next.

II. Methods

We evaluate the impact of reducing photon flux rates in the CT data on the reconstructed PET image and CT radiation dose. We used Catsim [4] for the simulation of the CT images and estimation of the CT dose. For the PET images we used a simplified version of ASIM [5], where the output of Catsim was used as the attenuation correction data for PET simulations as illustrated in Figure 2. Both Catsim and ASIM are analytic simulation methods. In other words they do not use photon tracking, but rather calculate noiseless line-integrals, followed by addition of statistical noise. The advantage of this approach is that multiple i.i.d. realizations of data can be rapidly generated, while the disadvantages include more complexity in modeling physics effects.

CT noise was generated using the following model of Equation (1) [4,13,18], where the electronic noise was adjusted so that it comprised 0.1% of the total noise for a CT technique of 50 mAs. The CT simulation includes effects of the x-ray tube spectra, beam conditioning, detector noise, and beam hardening correction. The PET simulation includes the effect of attenuation and photon counting.

$$y_i = G_i \sum_k E_k \cdot \text{Poisson} \left\{ \text{DQE} \cdot A_{ik} \cdot \sum_s \frac{1}{S} \exp \left(- \int_{L_i} \mu(x, E_k) dl \right) + s_{ik} \right\} + \text{Normal} \{d_i, \sigma_e^2\} \quad (1)$$

The test object used was a uniform 20×30 cm elliptical cylinder containing FDG in water at a typical clinical concentration. We evaluated bias and noise in the CT and PET images from 25 realizations.

The PET/CT system modeled with the GE Discovery STE, although only a single trans-axial section of the phantom equivalent to a single CT slice was evaluated. From this arrangement we evaluated the (1) radiation dose, (2) CT image noise and bias, (3) PET image noise and bias. As discussed by Colsher et al [7] we also evaluated the effect of a noise suppression method: Using simple box-car smoothing of the sinogram raw data, followed by additional smoothing in sinogram rows using the adaptive trimmed mean (ATM) algorithm [6] to further suppress noise in the raw CT sinogram before log-conversion and beam hardening corrections.

The CT techniques and parameters were as follows: Tube voltages 80, 100, 120, and 140 kVp. Rotation time 0.5 s. Tube current 500 down to 0.1 mA. Slice thickness 5 mm. Reconstruction 128×128 pixels over a 50 cm FOV (to match PET image dimensions). For the CT acquisition there was no bowtie filter and beam conditioning included 2 mm of graphite and 0.25 mm of aluminum. The PET images were reconstructed using filtered back projection using 128×128 pixels over a 50 cm FOV.

The CT images were converted to linear attenuation coefficients at 511 keV using a modification of the bilinear scaling method [16,17]. The conversion parameters used for the

CT-based attenuation correction (CTAC, Figure 2) are illustrated in Figure 3. These are based on those used in our clinical GE DSTE PET/CT scanner.

III. Results

A. Radiation Dose from the CT scan

Figure 4 presents the estimated radiation dose at 50 mAs for the four tube potentials for one realization of the CT scan simulation. The radiation dose was estimated using the Catdose component of Catsim (Figure 2). The total estimated dose increased as $(kVp)^{2.3}$ ($R^2=1.00$).

Using the results of Figure 4, we also estimated the relationship between the kVp and mAs on total dose, which is plotted in Figure 5, using $\log(\text{dose})$.

B. CT Image Noise and Bias

Figures 6 and 7 present the measured bias and noise in the CT images measured in each realization using 18 ROIs placed inside ellipse, each with a diameter of 1.5 cm. Results were calculated as a function of mAs, kVp and whether or not smoothing is applied to the sinogram. As expected there is a mAs threshold below which the CT bias and noise increase significantly.

We also observe that regardless of the acquisition and/or processing methods used, there is a critical level where approximately 0.1% of the sinogram elements were non-positive. This is illustrated in Figure 8. Above this threshold the noise and bias artifacts increased rapidly with decreasing mAs.

To further understand the relationship between dose and CT bias and noise effects, we plotted the percentage of non-positive sinogram elements as a function of mAs and kVp (Figure 9), analogous to the contour plot shown in Figure 5.

C. PET Image Noise and Bias

Figures 10 and 11 present the measured bias and noise in the PET images as a function of the mAs and kVp of the CT image acquisition. The effect of CT sinogram smoothing is also shown.

Given the wide ranges of bias and noise in the PET images, we also evaluated the range of PET image noise up to a specified limit of $\pm 5\%$ bias in the PET image. The resulting noise levels are shown in Figure 12, which indicate that the increase in PET image noise is typically less than 4% in this regime.

Discussion

The results presented here are based on simulation studies, but indicate promising directions. The first is that PET noise is not unduly increased by ultra low dose CT scans with the proviso that bias in the PET image is within an acceptable range. The second, and more important, indication is that there is significant room to reduce CT image dose in the cases where CT is not used for diagnostic purposes. This may occur, for example with respiratory-gated PET studies of the thorax. Table 1 lists several methods to reduce CT dose, while Table 2 lists a set of methods that can be employed to reduce the commensurate increases in noise and/or artifacts.

There are no basic physics limitations to reducing the radiation dose from extended continuous CT scans. There are, however, significant challenges due to the requirement of the CT scanner to provide 'diagnostic' CT images, with sufficient image quality to perform

a range of tasks from detection to estimation of subtle disease. It is well known that even within the available range of CT techniques it is possible to produce CT images that will bias the results of CT-based attenuation correction. Thus dose reduction methods that would compromise the diagnostic capability of a CT scanner to enable ultra low dose CT imaging, e.g. by replacing the CT tube with a low current version or by altering the data acquisition system (DAS), are not considered feasible.

There are several challenges to this process, specifically that diagnostic CT x-ray tubes cannot operate at very low current (mA) levels. An analogy is a high-performance sports car, which can drive at 100 mph that is then asked to drive at precisely 0.01 mph. Even if the x-ray photon flux could be lowered to reduce patient dose, artifacts results from the detectors and DAS being designed for high-flux and high-resolution operation. For PET attenuation correction at 511 keV, however, the tolerance for relatively high noise, low resolution, and low contrast [12,16] allows for the development of new methods to suppress noise and artifacts resulting from low x-ray photon fluxes. The methods discussed here may also enable other low-dose CT acquisition methods, e.g. dynamic contrast-enhanced CT imaging [3], longitudinal studies, or evaluation of therapy.

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Figure 1.
PET/CT data processing, indicating impact of CT image on PET data.

Figure 2.
Data processing flow used for the simulation studies.

Figure 3.
Conversion of CT image values to linear attenuation coefficients for CT-based attenuation correction (CTAC).

Figure 4. Radiation dose distribution for a 20×30 cm elliptical water cylinder as a function of tube voltage (kVp) at 50 mAs 5mm thick slice. Also shown is the estimated total radiation dose (mGy).

Figure 5.
Relative log(dose) contour plot as a function of mAs and kVp.

Figure 6.

Mean ($n = 25$) bias in the CT images as a function of mAs, kVp, and if smoothing is applied to the sinogram.

Figure 7. Ensemble ($n = 25$) noise in the CT images as a function of mAs, kVp and if smoothing is applied.

Figure 8. Relationship between the bias in the CT images and the percentage of non-positive sinogram elements.

Figure 9.
Percentage of non-positive sinogram elements as a function of mAs and kVp.

Figure 10.

Mean ($n = 25$) bias of PET images as a function of CT mAs, kVp, and CT sinogram smoothing. The true PET value is 20.2 (a.u.).

Figure 11.

Ensemble noise ($n = 25$) of PET images as a function of CT mAs, kVp, and CT sinogram smoothing.

Figure 12.

Same results as Figure 11. But clipped if PET image bias exceeds $\pm 5\%$.

Table 1

Methods proposed for reducing CT radiation dose for CT-based attenuation correction.

Reduce mAs
Optimize kVp
X-ray tube pulsing
Filtration and/or beam conditioning [10,11]
Collimation
Short scan

Table 2

Potential methods for CT noise / artifact suppression.

Reduced requirements for PET CTAC [12,16]
Combining detector elements
Sinogram smoothing/denoising (e.g. [7,13])
Iterative reconstruction (e.g. [14])
Compressed sensing (e.g. [15])
Image smoothing/denoising
Time averaging