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Noise and Bias Properties of Monoenergetic Images from DECT used for Attenuation Correction with PET/CT and SPECT/CT

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Abstract

We evaluate the energy dependent noise and bias properties of monoenergetic images synthesized from dual-energy CT (DECT) acquisitions used to estimate attenuation coefficients at PET or SPECT energies. This is becoming more relevant with the increased use of quantitative imaging by PET/CT and SPECT/CT. There are, however, variations in the noise and bias properties of synthesized monoenergetic images as a function of energy. We used analytic approximations and simulations to estimate the bias and noise of synthesized monoenergetic images of a water-filled cylinder from 10 to 525 keV. The dual-energy spectra were based on the GE Lightspeed VCT scanner at 80 and 140 kVp. Both analytic calculations and simulations for increasing energy the relative noise plateaued near 140 keV (i.e. SPECT with ^{99m}Tc), and then remained constant with increasing energy up to 511 keV and beyond (i.e. PET). If DECT is being used for attenuation correction at higher energies, there is a noise amplification that is dependent on the energy of the synthesized monoenergetic image of linear attenuation coefficients. For SPECT and PET imaging the bias and noise levels of DECT based attenuation correction is unlikely to affect image quality.

Keywords

dual-energy CT; attenuation correction; PET; SPECT

1. Introduction

In quantitative imaging by positron emission tomography (PET) and single-photon emission computer tomography (SPECT), correction for the effect of attenuation is of paramount importance. At the energies of X-ray CT, attenuation is due to Compton scatter and photoelectric absorption, while at SPECT energies, and particularly PET energies, Compton scatter is the dominant process for biological materials [1]. PET/CT and SPECT/CT scanners are used to provide accurately aligned functional and anatomical information [2]. A secondary synergy of dual-mode PET/CT and SPECT/CT scanners is to use the CT image for attenuation correction of the PET emission data [1, 3]. A list of energies of interest are given in Table 1, while Figure 1 plots the mass attenuation coefficients of common materials over the energy ranges relevant to PET/CT and SPECT/CT imaging.

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CT-based attenuation correction (CTAC) offers several benefits over transmission scanning with an isotope at or near the energy of the nuclear tracer isotope. CTAC provides low-noise attenuation correction factors without the need for lengthy nuclear transmission scans, significantly reducing the overall scan time. Moreover, CT scans contain no bias from the emission contamination present in post-injection imaging.

CT attenuation correction factors are susceptible to bias due to the need to transform the CT image to attenuation correction factors at the nuclear isotope energy, typically either at 140 or 511 keV. Three potential solutions are employed for transforming attenuation images from CT energies to SPECT and PET energies. Segmentation methods can be used to separate the CT image into regions corresponding to different tissue types, which then are replaced with appropriate attenuation coefficients at the appropriate energy (e.g. 140 or 511 keV). This method has a potential source of error due to tissue misclassification. Secondly, linear scaling of the entire CT image with the ratio of attenuation coefficients of water (representing soft tissues) at the photon energies of CT and the target energy offers a simple solution for the transformation. For bone, however, linear scaling is a poor approximation, since photoelectric absorption dominates Compton scatter at the lower range of CT energies [4]. Finally, the most common method for transforming to PET energies is Bilinear/Hybrid Scaling. In these methods, different scaling factors (for water and air, and for water and bone respectively) are used to calculate the attenuation values for CT numbers H for which $-1000 < H < 0$, and for $H > 0$ [4-6]. The bilinear scaling method and other hybrid methods have been shown to give reasonable results for low- Z biological materials in practice. However, for high- Z materials such as contrast agent there is possibility for significant bias [1].

Dual energy CT was proposed as a method to estimate material properties using either photoelectric and Compton components [7] or physical basis materials such as plastic and aluminum [8]. In addition, DECT has been proposed to remove the bias from the CTAC image for SPECT [3, 6] and PET [9, 10]. DECT is problematic because of the significant noise amplification and the additional patient radiation dose required to perform two scans and to reduce noise. To address this we have been investigating alternate methods using iterative CT reconstruction [11] and image-based strategies [10]. In these study we evaluate the energy dependent noise properties of monoenergetic images synthesized from dual-energy CT, as the behavior in the energy range appropriate for nuclear imaging 140-511 keV) varies from the X-ray mean energy behavior (~ 75 keV).

2. Methods

The general procedures of the dual energy method are outlined in Figure 2. The methods of separating the dual-kVp sinogram data via the basis material decomposition (BMD) technique [8] have been well described. Here we note that we are interested in the bias and noise of the monoenergetic attenuation image $\mu(E)$ at energies ranging from 140 to 511 keV.

We first used analytic approximations to estimate the variance of the synthesized monoenergetic sinogram. This approach was based on the work of Doost-Hoseini [12], where the variance of the monoenergetic sinogram is approximated by

$$\sigma_m^2 = \frac{(f_1(E_m)m_{22} - f_2(E_m)m_{21})^2 \frac{\sigma_1^2}{I_1^2} + (f_1(E_m)m_{12} - f_2(E_m)m_{11})^2 \frac{\sigma_2^2}{I_2^2}}{(m_{11}m_{22} - m_{12}m_{21})^2}, \quad (1)$$

where I_1 and I_2 and σ_1^2 and σ_2^2 are the integrated photon signals and variances for the two kVp scans, $f_1(E)$ and $f_2(E)$ are the energy dependence of the attenuation coefficients of the two basis materials, E_m is the target energy and $m_{ij} = (I_i/I_j)(A_i/A_j)$ is the variation of signal intensity with amount of basis material. This derivation assumes monoenergetic x-rays and thus ignores beam hardening effects, so we also used simulations to estimate the noise of the synthesized monoenergetic data from 10 to 525 keV.

We used the CatSim package for simulation of the CT images and estimation of the CT dose [13]. The CT simulations included the effects of the x-ray tube spectra, beam conditioning, and quantum noise. The effects of bowtie filters were not included to avoid having to repeat the calibration for each fan angle. The dual-energy spectra were based on the GE Lightspeed VCT scanner at 80 and 140 kVp, with added filtration of 0.5 mm Cu. We used the basis material decomposition (BMD) technique [8] with basis materials of aluminum and plastic (polyethylene). For calibration we defined plastic and aluminum step wedge phantoms (Figure 3).

Calibration (Figure 3)

1. Define step wedges made of aluminum and plastic for calibration (Al: 0-8 steps, 4mm/step; Plastic: 0-17 steps, 12.5mm/step) as shown in Figure 3.
2. Generate sinograms from high and low kVp scans for the step wedges.
3. Based on the sinograms, obtain calibration tables of $\ln(I/I_0)$ for each combination of aluminum and plastic steps.
4. Using Basis Material Decomposition (BMD technique), generate calibration coefficients $\{d_i\}$, $\{e_j\}$ for each material:

$$A_\alpha = d_0 + d_1 T_H + d_2 T_L + d_3 T_H T_L + d_4 T_H^2 + d_5 T_L^2 + d_6 (T_H T_L)^2 + d_7 T_H^3 + d_8 T_L^3 \quad (2)$$

$$A_\beta = e_0 + e_1 T_H + e_2 T_L + e_3 T_H T_L + e_4 T_H^2 + e_5 T_L^2 + e_6 (T_H T_L)^2 + e_7 T_H^3 + e_8 T_L^3 \quad (3)$$

where A_α and A_β are the thickness of the two materials, and $T_H = \ln(I_0/I)_H$ and $T_L = \ln(I_0/I)_L$ are the log ratios for the high (H) and low (L) kVp scans.

Measurements (Figure 2)

5. Generate high and low kVp sinogram data for the water phantom at the given settings (mAs).

6. Sinogram decomposition: Using generated coefficients $\{d_i\}$, $\{e_j\}$ from Step 4 above, and measured data from two kVp scans from Step 5, obtain basis material sinograms of A_α and A_β for the phantom.
7. Reconstruct basis material density images.
8. Convert density images to attenuation images, at desired energy E , using known attenuation values versus E for each material.
9. Sum attenuation images to obtain the total attenuation at the desired energy E .

The simulations used two different approaches. In the first approach we simulated a 10 cm diameter water cylinder and looked at the variance in the central projection bin of the synthesized monoenergetic sinograms, which assumed that the noise is uncorrelated in this manner. This allowed rapid calculation of the variance at every integer value of keV. In the second approach we evaluated the noise and bias in reconstructed images of a 20×30 cm elliptical water cylinder that was 10 cm long axially. Due to the long computational time required, only selected keV energies (e.g. Table 1) were used. In this case the noise and bias were evaluated from 18 small circular ROIs placed on the reconstructed images, where the mAs of both the high and low kVp scans were varied. As a check, we compared the linear attenuation coefficients of the synthesized monoenergetic images from simulations of the entire DECT process with known true values [14]. Finally we also estimated the total dose to the 20×30 cm water phantom as a function of the mAs of the high and low kVp scans

3. Results

The pre- and post-filtration spectra are shown in Figure 4. Characteristics of the spectra are listed in Table 2.

The coefficients $\{d_i\}$, $\{e_j\}$ describing the relationships between aluminum and polyethylene thickness and the high- and low-kVp log-transformed projections (equations (2) and (3)) were estimated as:

$$A_{Pl} = -0.0482 + 22.6207T_H - 15.4241T_L + 1.9393T_H T_L + 0.9391T_H^2 - 2.5295T_L^2 - 0.0107T_H^2 T_L^2 - 0.2456T_H^3 + 0.2268T_L^3 \quad (4)$$

$$A_{Al} = 0.0160 - 4.4844T_H + 4.1199T_L - 1.3491T_H T_L + 0.0590T_H^2 + 1.1774T_L^2 + 0.0032T_H^2 T_L^2 + 0.0792T_H^3 - 0.0715T_L^3 \quad (5)$$

For the cylinder water phantom of diameter 10cm the relative noise as a function of energy is shown in Figure 5, with key values listed in Table 3. The bias of linear attenuation coefficients of the synthesized monoenergetic sinograms were within 1% of the known true values across the energy range evaluated (Figure 6).

Relative noise at a fixed mAs for the high and low kVp scans of a 20×30 cm elliptical water cylinder is shown in Figure 7. There are two major differences in the calculations used for Figure 7. First, the noise was estimated from reconstructed images of a 20×30 cm elliptical water cylinder. Second, the noise was converted to a relative noise by dividing by

the attenuation coefficient for water at each energy (Figure 1). For comparison purposes we also show the analytical approximation (Equation (1) and Figure 5) normalized by the attenuation coefficient. In this case we see that the relative noise plateaus at a level approximately 2.5 times the lowest level at the optimal energy. In addition we note that the analytic approximation provides a close estimate.

The effect of the mAs for high and low kVp scans on noise and bias in the monoenergetic image was evaluated at the fixed energies indicated in Figure 7. Figure 8 shows the noise and bias at 511 keV. For energies from 140 to 511 keV the dependence of noise and bias on the mAs for high and low kVp scans had similar patterns.

The radiation dose to the 20×30 cm diameter water phantom as a function of mAs for high and low kVp scans is shown in Figure 9, demonstrating the expected inverse relation to noise (Figure 8). Also shown is a plot of $1/(\text{noise} \times \text{dose})$ as a function of mAs for high and low kVp scans, which indicates that lower mAs levels are desirable from an SNR/dose ratio measurement. However, lower mAs levels will be limited by unacceptable levels of bias (Figure 8: Right).

4. Discussion and Conclusions

Standard methods of CT-based attenuation correction for SPECT and PET have been shown to work well for low density biological objects [1]. In cases where there is PET or SPECT tracer uptake in bone or other confounding high-Z materials (e.g. contrast agent, implants) more accurate methods such as dual-energy CT-based attenuation correction may provide significantly more accurate images [3,10]. To analyze the properties of DECT-based attenuation correction we focused on the noise and bias of synthesized monoenergetic attenuation image as a function of the synthesized energy and source currents. Although there are advantages for using different filtration for the two sources and a dual-tube scanner, we used the same filtration for both kVp scans as a single-tube CT scanner is more likely to be combined with a SPECT or PET scanner. The radiation doses presented above (Figure 9) were calculated without the use of a bowtie filter. However, we repeated the dose estimates with a standard bowtie filter in place and the same trends as in Figure 9 were observed.

Analytic calculations and simulations showed the expected minimum noise value for a synthesized monoenergetic image at an energy between the mean energies of the two source spectra. In addition we found that for increasing energy the relative noise plateaued near the 140 keV energy of $^{99\text{m}}\text{Tc}$ (i.e. SPECT), at a level of approximately 2.5 times the minimum relative noise, and then remained constant with increasing energy (i.e. at PET imaging energies). The analytic calculations and simulations do not agree completely, possibly due to the assumption of monoenergetic source spectra. However, the analytic approximations of equation (1) can provide a useful guide to noise behavior. As a check, the bias of the linear attenuation coefficients of the synthesized monoenergetic images were within 1% of the known true values across the entire energy range.

Analysis of the noise, bias, and radiation dose indicated that there are optimal ranges for the mAs of the high and low kVp scans. Within a normal clinical range of mAs levels,

however, the noise and bias of DECT based attenuation correction are unlikely to affect SPECT or PET image quality. However, the precise impact of DECT based attenuation correction on SPECT and PET image noise and bias remains to be determined.

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Figure 1. Mass attenuation coefficients of materials of interest over energy ranges relevant to PET/CT and SPECT/CT imaging [14].

Figure 2.

Process used to form the monoenergetic attenuation image $\mu(E)$ for attenuation correction of a SPECT or PET radioisotope emitting photons with and energy E .

Figure 3. Schematic illustration and simulation of x-ray image of step wedge used for BMD calibration.

Figure 4.
 Left: Incident spectrum before any filtration. Right: Incident spectrum after filtration and before being attenuated by imaging object.

Figure 5. Noise vs. synthesized energy of monoenergetic image for approximate theoretical prediction (Equation (1)) and simulation results for a 10 cm diameter water phantom. Left: Overview with log scale. Right: Detailed comparison using linear scale and with energies for TC-99m and F-18 indicated.

Figure 6.
Percentage error of linear attenuation coefficient of water obtained from simulation of complete DECT process to form monoenergetic image.

Figure 7.

Relative noise vs. synthesized energy of monoenergetic image for theoretical prediction (divided by attenuation coefficient) and simulation results based on reconstructed images at indicated energies for a 20×30 cm water phantom.

Figure 8.

Noise (left) and bias (right) in the 511 keV monoenergetic image as a function of tube current mAs for high vs. low energy (kVp) scans. Contour levels and color bars are in relative units.

Figure 9.

Left: Total radiation dose (mGy) to 20×30 cm diameter water phantom as a function of mAs for high vs low energy (kVp) scans. Right: Contour plots of $1/(\text{noise} \times \text{dose})$ as a function of mAs for high vs low energy (kVp) scans. Contour levels and color bars are in relative units.

Table 1

Energies of common isotopes used in PET/CT and SPECT/CT Imaging

Mode	Isotope	Energy (keV)
PET	All (i.e. F-18, C-11, N-13, etc.)	511
SPECT	Xe-133	80.9
SPECT	Tc-99m	140
SPECT	I-123	160
SPECT	Ga-67	185
SPECT	In-111	171, 245

Table 2

Mean energy before and after filtration for the two kVp scans in single tube

Spectrum	80 kVp	140 kVp
Pre filtration	42.7 keV	59.7 keV
After filtration (0.5mm Cu)	55.6 keV	74.3 keV

Table 3

Comparison of noise ratios from theoretical prediction and simulations shown in Figure 5

Method	Minima (energy)	140 keV	511 keV
Theoretical (eqn 1)	1.04 (69 keV)	2.71	1.36
Simulation	1.02 (72 keV)	2.54	1.33