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Correction technique for cascade gammas in I-124 imaging on a fully-3D, Time-of-Flight PET Scanner

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Abstract

It has been shown that I-124 PET imaging can be used for accurate dose estimation in radio-immunotherapy techniques. However, I-124 is not a pure positron emitter, leading to two types of coincidence events not typically encountered: increased random coincidences due to non-annihilation cascade photons, and true coincidences between an annihilation photon and primarily a coincident 602 keV cascade gamma (true coincidence γ -ray background). The increased random coincidences are accurately estimated by the delayed window technique. Here we evaluate the radial and time distributions of the true coincidence γ -ray background in order to correct and accurately estimate lesion uptake for I-124 imaging in a time-of-flight (TOF) PET scanner. We performed measurements using a line source of activity placed in air and a water-filled cylinder, using F-18 and I-124 radio-isotopes. Our results show that the true coincidence γ -ray backgrounds in I-124 have a uniform radial distribution, while the time distribution is similar to the scattered annihilation coincidences. As a result, we implemented a TOF-extended single scatter simulation algorithm with a uniform radial offset in the tail-fitting procedure for accurate correction of TOF data in I-124 imaging. Imaging results show that the contrast recovery for large spheres in a uniform activity background is similar in F-18 and I-124 imaging. There is some degradation in contrast recovery for small spheres in I-124, which is explained by the increased positron range, and reduced spatial resolution, of I-124 compared to F-18. Our results show that it is possible to perform accurate TOF based corrections for I-124 imaging.

I. INTRODUCTION

Accurate dose measurement over an extended period of time is necessary to properly assess the dose delivery in radio-immunotherapy, and PET due to its high sensitivity and spatial resolution is the preferred in vivo imaging modality in these situations. The I-124 positron emitter which has a relatively long half-life (4 days) is useful for dose monitoring over an extended time period [1–6]. The I-124 radio-nuclide is not a pure positron emitter and has a cascade of γ emissions. The branching ratio for positron emission is 0.23 and about 50% of the time the positron emission is immediately followed by a 602 keV γ emission. Besides the complicated decay scheme of I-124 which gives rise to cascade γ 's, the positron range is also larger compared to F-18, leading to a degradation in reconstructed spatial resolution of about 1-mm [7].

The low branching ratio for positron emission translates into reduced sensitivity for PET imaging. Since TOF PET imaging has been shown to improve F-18 image quality with effective gains in image signal-to-noise ratio leading to shorter scan times [8–11] compared to Non-TOF PET, it will be desirable to use TOF PET to also achieve good quality I-124 images in order to compensate for the low positron emission branching ratio. However, several factors related to the emission of cascade γ 's can lead to problems in generating quantitative images, especially in TOF modes. The energy of these γ emissions is very close to 511 keV and the energy resolution in the current generation of PET scanners is not sufficient to reject them via energy gating. As a result the collected coincident events in the PET scanner will be comprised of:

- True positron annihilation coincidences (T)
- Scattered positron annihilation coincidences (Sc)
- Random coincidences due to photons from annihilation events (R)
- Random coincidences due to one or both photons coming from the cascade γ 's (R_c)
- True coincidences between one photon from an annihilation event and the other photon primarily from a coincident 602 keV cascade γ (true coincidence γ -ray background, or T_c)

The R coincidences are measured using a delayed window technique in both Non-TOF (radial distribution) and TOF PET (radial and time distribution). In I-124 TOF PET imaging the delayed window estimate will also provide an estimate (in both radial and time directions) of $R+R_c$, since it measures coincidences between all single photons hitting the detector. In the past, several investigators have evaluated different techniques for subtracting the T_c coincidences arising from imaging non-pure positron emitters in Non-TOF PET scanners. These ranged from (i) convolution subtraction techniques in 2D PET scanners for subtracting both the T_c coincidences as well as the Sc coincidences [12–14], through (ii) convolution subtraction for Sc coincidences followed by a linear tail fitting estimate in the sinograms for T_c coincidences in 2D scanners [15], to (iii) uniform [16,17] or linear [18] tail fitting estimate in the sinograms for T_c coincidences followed by a standard Sc correction technique in either 2D or fully-3D PET scanners. Most modern PET scanners operate exclusively in the fully-3D mode where the convolution subtraction technique for Sc coincidences has limitations in accurately estimating the scatter [19]. In these scanners, the Sc coincidences are estimated (within 5–10% depending upon patient size) using the model-based Single Scatter Simulation (SSS) [20–22] where the tails in the sinogram are used to scale the scatter estimate to the absolute number of scattered events. This introduces a problem when imaging non-pure positron emitters and using a tail-fitting procedure to first estimate and subtract the T_c coincidences (as in technique (iii) above), since the sinogram tails now include Sc as well as T_c events. The fraction of Sc to T_c events is not a fixed constant as assumed in the past [16–18], since the Sc events are a function of the patient size. In addition, as has been pointed out recently [23], an estimate of the time distribution of the scattered coincidences is needed when performing TOF image reconstruction. For TOF PET, the SSS technique has been successfully extended to include timing information and thus provides the scatter estimate in radial and time directions [23,24]. A primary aim of this investigation is, therefore, to characterize the radial and time distributions of the T_c coincidences. This is followed by a modification of our TOF-extended SSS estimation algorithm to build in the estimation of Sc and T_c coincidences all in one step, and achieve high quality, and quantitative I-124 TOF PET images. The benefit of TOF imaging over Non-TOF imaging in fully-3D PET has already been demonstrated in other publications [8–11] (for F-18 imaging), while cascade γ correction techniques have also been successfully implemented for Non-TOF scanners [12–18]. As a result, in this paper the focus is only on developing a true coincidence γ -ray background correction technique applicable to TOF PET imaging.

II. SCANNER DESCRIPTION

All measurements were performed on the Gemini TF PET/CT (Philips Medical Systems, Highland Heights, OH) scanner [25] which is a new high performance, fully-3D TOF PET scanner combined with a 16-slice Brilliance CT scanner. The PET component of the Gemini TF is comprised of 28 flat modules, each a 23×44 array of $4 \times 4 \times 22\text{-mm}^3$ LYSO crystals. The scanner ring diameter is 90.34-cm. Annular lead shielding (1-inch thick) at the two axial ends is used to reduce detection of events occurring outside the field-of-view (FOV). The patient bore has a diameter of 71.70-cm with active transverse and axial FOVs of 576 and 179.8-mm, respectively. This scanner has a measured spatial resolution of 4.8-mm near the center of the scanner and an intrinsic timing resolution of 585ps at low count-rates. The energy resolution for this scanner is 12% at 511 keV leading to a default energy window of 440–665 keV. All measurements in this investigation were performed with this default energy window. In I-124 imaging, multiple coincidences (i.e. detection of more than two photons within the coincidence window) due to the emission of cascade γ 's will be a significant effect that will vary depending upon the scanner electronics. In Gemini TF, the electronics only collect up to two valid pairs of coincident events in a given clock cycle (10ns). If there are more than two coincidences (within the coincidence window of ± 3 ns of each other) in the clock cycle, then all events are rejected. So for triple single events in the clock cycle, there will be some situations where one or two valid coincidences are written out, while at other times (each single in coincidence with the other two, i.e. three valid coincidences) no events are collected.

III. CHARACTERIZATION OF RADIAL AND TIME PROFILES

A. Radial profiles for a line source

A 20-cm diameter by 19-cm long NEMA NU 1994 [26] line source scatter cylinder was used to characterize the radial and time profiles for F-18 and I-124 positron emitters. The line source is made of glass with an inner diameter of about 1-mm and wall thickness of 1-mm as well. The line source was filled with activity and placed at the center of the cylinder. The cylinder was left empty for initial measurements and was approximately centered in the scanner. The activity in the line source was 11.5 and 15.5-MBq (0.31 and 0.42-mCi) for the F-18 and I-124 measurements, respectively. Fig. 1 shows the radial profiles for the two data sets. Since the line source was not perfectly centered in the scanner, a procedure analogous to making profiles for scatter fraction estimation in the NEMA NU 1994 standards was used. Briefly, in each sinogram slice and for each projection (in-plane) angle the pixel with maximum counts is determined. Every projection angle within this slice is then shifted so that the pixel containing the maximum counts is aligned with a pre-defined central pixel (radial bin of 256 in this work). In this manner a radial profile is drawn for every sinogram slice. These profiles are for raw sinogram counts without any corrections and as shown are summed over all slices in the central 4-cm length of the line source. The I-124 radial profile (Fig. 1B) has a non-zero, flat offset after subtraction of delayed counts from the prompts. Variation of the radial profile more than 50 bins (10-cm) from the central peak is $< 10\%$. This flat offset in the radial profile shows that the distribution of true coincidence γ -ray background events is uniform in the radial direction, and represents about 23% of the prompt-delayed counts (from an analysis based upon NEMA NU2 2001 scatter estimate technique [27]). The slight asymmetry in the I-124 profile is due to a small off-center position of the line source when placed in the scanner.

The line source measurements were then repeated after filling the cylinder with water. The water background now adds additional scatter in the collected data and represents a more realistic clinical imaging situation. The activity in the line source for these measurements was 10.0 and 15.5-MBq (0.27 and 0.42-mCi) for F-18 and I-124, respectively. Fig. 2 shows the radial profiles as measured for the two isotopes. Compared to the profile shown in Fig. 1A, there is now a slowly varying non-uniform offset in the F-18 data set (Fig. 2A). These counts

represent the radial distribution of the scattered coincidences arising due to the presence of the water background present in the cylinder. The I-124 profile (Fig. 2B), on the other hand, has the additional flat offset arising from the true coincidence γ -ray background as seen earlier for the line source in air measurement (Fig. 1B). The scatter fraction as measured with the F-18 data is 23% and increases to 46% with I-124, which now includes scatter and true coincidence γ -ray background events.

B. Time profiles for a line source

The data collected with the line source were also binned into timing histograms. Fig. 3 shows the normalized time profiles for line source in air (Figs. 3A and 3C) and line source in water (Figs. 3B and 3D) measurements. The timing bins represent the difference in arrival times of the two photons in the detectors and each bin is 25 ps in size. The figures were normalized to peak counts in each data set in order to show the difference in shapes for F-18 and I-124 data. The profiles are shown for prompts-delayed coincidences. On a linear plot (Figs 3A and 3B) the difference between the F-18 and I-124 is not very clear, but the time profiles for I-124 are a little wider (3 bins at the FWHM level and 13 bins at the FWTM level) than those for F-18 when plotted on a logarithmic scale (Figs 3C and 3D). The slightly wider time profile for I-124 is most likely due to those coincident events where one of the two contributing photons is a cascade γ (T_c). This indicates that, as a first approximation, the true coincidence γ -ray background can be approximated as a flat radial profile and a time profile that is the same as that for the scattered events. In our TOF-extended single scatter simulation we first model the scanner using a sparsely modeled subset of uniformly placed crystals [24]. The attenuation image is then uniformly distributed with scatter points, including points outside the FOV. For each line-of-response (LOR) the contribution of scatter events is then calculated as a weighted sum of events arising from each of these scatter points. For each of these scatter points, the emission point can lie anywhere along the paths of the two photons. The weight is a product of several factors including the probability of interaction at the scatter point, the attenuation along the segments of the path, the probability of scatter through the required angle (from Klein-Nishina cross-section), and the solid angle of the two detectors as seen from the scatter point. For TOF extension, we also calculate the arrival time difference between the two photons contributing to the scatter event by computing the distance traveled from the emission point to entry at the detector plane. As a final step, the radial profile of the scatter estimate without TOF information is then scaled as a polynomial fit to the tails of the radial profile for emission data. For corrections of the I-124 data we, therefore, add a constant to this polynomial fit procedure performed between the scatter estimate and emission data radial profiles. While addition of an offset value to tail-fitting in standard scatter correction techniques has previously been mentioned and used [16], our technique extends its use in combination with a TOF scatter correction method for reconstructing TOF PET data.

IV. IMAGE QUALITY

Imaging data were acquired using a cylindrical phantom 20-cm in diameter and 19-cm long. The cylinder contained one 3.0-cm diameter, one 1.5-cm diameter, and three 1.0-cm diameter spheres representing varying lesion sizes. Data were first acquired using F-18 followed by a measurement using I-124 radioisotope. Both measurements were performed with a total activity of 54.4-MBq (1.47-mCi) with the activity concentration ratio in the spheres being 6:1 with respect to the cylinder background. All data sets were corrected for attenuation (using CT-based attenuation factors) and scatter (using the model based single scatter simulation). The I-124 data were reconstructed both with and without the offset based tail-fitting of the radial profiles of the scatter estimate obtained through the TOF-extended single scatter simulation. In Fig. 4 we show the TOF reconstructed images for a 5-minute F-18 scan, as well as 5 and 20-minute I-124 scans where images with and without the offset correction for true

coincidence γ -ray background are shown. Comparing the two sets of I-124 images, it is very clear that without the offset correction for true coincidence γ -ray background, there is a non-uniformity (lower counts at the center of the cylinder) present in the image that will translate into incorrect quantitation. This non-uniformity arises due to the tail-fitting procedure in the scatter estimation process. For I-124 data, if no constant offset is used in the tail-fitting step, then the algorithm overestimates the scatter in the center of cylinder as it tries to fit a polynomial function to high tails (due to constant true coincidence γ -ray background). Consequently, the reconstructed image has low intensity at the center as seen in Figs. 4D and 4E. With offset correction, the 5-minute I-124 scan, while uniform looking, is noisier than the corresponding F-18 image. This is explained by the reduced branching ratio (0.23) for positron emission in I-124 decay. The 20-minute I-124 scan, however, looks very similar in quality to the 5-minute F-18 scan. This four-fold increase in scan time compensates for the reduction in number of positron emissions for the same activity amount of I-124 activity as F-18.

In Table I we show the measured contrast recovery coefficients (CRC) calculated in a manner analogous to that prescribed in NEMA NU2-2001 standard [27]. Circular regions-of-interest (ROIs) were drawn over each sphere equal in diameter to the physical diameter of the sphere to calculate the mean counts in the sphere region (C_H). Mean background counts (C_B) were estimated by drawing a 3-cm diameter circular ROI near the center of the transverse slice. The CRC was estimated using the formula:

$$\text{CRC} = 100 \times \frac{\frac{C_H}{C_B} - 1}{6 - 1} \quad (1)$$

The results shown in Table I are average results for measurements performed on three separate copies of 5 minute scan times each. The CRC values for the spheres in I-124 studies without offset correction are higher than those calculated for the F-18 data sets (greater than 100% for the 3.0-cm diameter sphere) due to over-subtraction of scatter counts at the center of the cylinder. With offset correction, the I-124 results are closer to the F-18 values. A paired Student t-test was performed for the CRC values measured with F-18 and I-124 after offset correction. The p-values for the 3.0, 1.5, and 1.0-cm diameter spheres were 0.02, 0.01, and 0.01, respectively, indicating that the differences in CRC values for the two data sets are statistically significant at > 98% confidence level. This reduction in CRC values after offset correction may be partially explained by the increased partial volume effect arising due to the increased positron range for I-124. As noted in [28], increased positron range in I-124 leads to a degradation in spatial resolution by 1.2-mm compared to F-18, which will lead to increased partial volume effects in I-124 imaging. For these measurements, we also estimated the relative pixel noise in the background region by normalizing the standard deviation of counts in the background region to the mean counts in the background region. The relative noise in the 5 minute F-18 scan was 9.9%, while the I-124 scan had a noise of 19.5 % for the same scan time. We also reconstructed I-124 data equivalent to a 20 minute scan time and measured a pixel noise of 11.5%. Defining noise equivalent counts (NEC) as

$$\text{NEC} = \frac{T \times T}{T + S_c + R_c + R} \quad (2)$$

we calculate an NEC of 41 Mcts and 31 Mcts for the 5 minute F-18 scan and 20 minute I-124 scan, respectively. Since the pixel signal-to-noise (SNR) is proportional to $(\text{NEC})^{1/2}$, the ratio of pixel SNR in the 5 minute F-18 scan to the 20 minute I-124 scan is therefore 1.15. This number is consistent with the ratio calculated from the measured pixel noise given above.

V. I-124 PET IMAGING IN THE PRESENCE OF I-131 BACKGROUND

In clinical imaging environment there are situations where a patient undergoes I-131 radiotherapy, and where the same compound first labeled with I-124 allows for accurate dose planning with a PET scan [1,29,30]. Using this approach, the I-124 labeled substance can subsequently also be used for dose evaluation with serial PET scans performed as the patient undergoes the I-131 therapy [2,3,6]. In these situations the I-124 PET scan will have a background of the I-131 radio-nuclide present in the data.

Since I-131 is single γ emitter with primary γ emissions of 637 keV (0.065 branching ratio), 364 keV (0.82 branching ratio), and 284 keV (0.06 branching ratio), there will be some contamination from these γ 's detected as random coincidences in the PET scanner. The lower energy gate on the Gemini TF scanner is set at 440 keV, so this contamination will be dominated by the higher energy γ 's (those at 637 keV). Since the patient is administered I-131 for therapy, the amount of I-131 present will be high leading to not only increased number of random coincidences, but also increased count-rate effects due to the elevated rate of single events in the scanner. The impact of increased single events and random coincidences on the NEC performance of an ECAT Exact HR+ scanner (CTI/Siemens) [31] has been previously studied [32]. The impact of I-131 background on I-124 imaging will, however, be a function of scanner design since it is determined by the scanner count-rate performance and the manner in which the electronics handle multiple coincidences. For our evaluation, we are more interested in studying the impact on I-124 CRC values as a function of increased I-131 background activity. In addition, since the Gemini TF represents the latest generation of PET scanners, it will be more representative of the impact of new scanner technology in I-124 imaging. We therefore performed an additional experiment to simulate the presence of varying amounts of I-131 background while imaging the I-124 lesion phantom. We filled the lesion phantom with a total of 40.7-MBq (1.1-mCi) of I-124 with a 6:1 activity uptake ratio between the spheres and the background. Another 20-cm in diameter by 19-cm long water filled (background) cylinder was placed end-to-end next to the I-124 lesion phantom. In a clinical setting, the distributions of I-124 and I-131 will be superimposed. However, since I-131 is a single γ emitter, its contribution to collected data will be in the form of random coincidences, which are uniformly distributed (as long as the activity is in the scanner FOV). Therefore, having the I-124 and I-131 sources physically separated in our experimental setup (as opposed to being superimposed) will not differ significantly from the clinical imaging environment in terms of the random coincidence distribution.

Data were collected for varying amounts of I-131 (0, 370, 740, and 1110-MBq, or 0, 10, 20, and 30-mCi) present in the background cylinder, with a 20 minute scan per I-131 activity level. The collected data were reconstructed after corrections were performed as outlined earlier. The singles rate in the scanner varied as 5.6, 16.0, 28.5, and 35.4 Mcps as the amount of I-131 present in the background cylinder increased. As shown previously [25], the corresponding timing resolution in the scanner at these singles rates is 610, 675, 750, and 815 ps as the singles rate increases. As a result, appropriate timing resolution values were used when reconstructing these data. Also, at singles rates of 28.5 and 35.4 Mcps the scanner is performing beyond the count-rates typical for ^{18}F -FDG oncology studies, and is susceptible to deadtime effects as well. In Fig. 5 we show representative slices from the reconstructed images for these four setups. The images with 740 and 1110-MBq (20 and 30-mCi) of I-131 show a non-uniformity in the background where the central region has reduced count-density, indicating an over subtraction of scatter and I-124 true coincidence γ -ray background events.

In Table II we show the results for CRC for the three sphere sizes and the relative pixel noise as measured for the four different levels of background I-131. The difference in the CRC values for I-124 in Table I and in Table II (without I-131 background) is due to the physical placement

of the phantom in the axial FOV. The two data sets were acquired on separate days. The data in Table I (both F-18 and I-124) were obtained at a fixed bed position where the sphere counts were split over three image slices (4-mm thick). The data in Table II which were acquired on a different day had the sphere counts split over only two slices, resulting in a better estimate of the CRC value. Unfortunately, this is one of the many practical imaging situations which affect the measured uptake in lesions leading to a variability in numbers for the same lesion size in different patients or different location within the same patient. The results from Table II, which were collected serially at the same bed position, indicate that for an I-131/I-124 ratio of at least 10:1, image quantitation is not affected while for a ratio of 20:1 and higher, we see increased CRC due to over-subtraction in the central region of the phantom.

VI. DISCUSSION AND CONCLUSION

I-124 imaging is an important PET imaging technique that will be utilized more as radio-immunotherapy procedures become more widespread. The quantitative imaging capability of PET and its high sensitivity makes it attractive for accurate estimation of dose delivery in these procedures compared to I-123 based single photon imaging procedures. Past research work has shown that it is possible to accurately correct for the true coincidence γ -ray background events that arise in I-124 imaging which is not a pure positron emitter. The low branching ratio for positron emission (0.23) also leads to poor image quality, which requires long scan durations to achieve high quality images. TOF PET with its increased effective sensitivity as shown in F-18 imaging also has the potential to reduce the scan time for these I-124 studies [8,9]. However, correction techniques for the true coincidence γ -ray background now have to account for the time distribution of these events as well. In this work we have used a TOF-extended SSS technique to correct for the I-124 true coincidence γ -ray background to achieve improved quantitative capability.

The results of this investigation show that the radial distribution of the true coincidence γ -ray background is uniform over the object size. Additionally, the timing distribution of these true coincidence γ -ray background events is also similar to the scattered coincidence events. Hence, the TOF-extended single scatter simulation can be used to estimate the scattered coincidences, as well as the true coincidence γ -ray background by adding a uniform offset in the radial dimension. In clinical human imaging, there is always some activity present outside the FOV, which contributes to additional scatter coincidences (out-of-FOV scatter). Also, in large size patients the number of multiple scatter events will increase. The tails of the emission data radial profiles are used in these situations to scale the SSS estimate for the multiple as well as out-of-FOV scatter events with a polynomial fit. In principle, the addition of an offset term to also scale for the I-124 coincidence γ -ray background can be incorporated as one procedure just as we did in this study. However, depending upon the patient size, this procedure may not work very well when the amount of multiple or out-of-FOV scatter increases significantly. This is a general limitation of the SSS scatter estimation procedure, and so the application of our I-124 correction technique will be valid for extended or larger source distribution only up to a certain limit, which was not a part of this investigation.

Imaging results with a small lesion phantom indicate that with these corrections, the CRC values in I-124 imaging are closer to the F-18 values. However, as measured by others [28], degraded spatial resolution in I-124 imaging (larger positron range) leading to partial volume effects, as well as some left over bias due to true coincidence γ -ray background, leads to reduced CRC values relative to F-18 imaging. For imaging situations where there is a presence of I-131 (single γ emitter) in the background, we have shown that the ability to perform quantitative I-124 imaging degrades once the relative amount of I-131 is between 10–20 times that of the I-124. These results show that with TOF PET, high quality I-124 imaging can be performed with accurate quantitative results for dosage estimation.

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Fig. 1.

Measurements of radial profiles (logarithmic scale) for a line source in air. (A) is for an F-18 source while (B) is for an I-124 source. Each radial bin corresponds to 2.09-mm. Note the flat offset present in the prompt-delays profile of I-124 due to the true coincidence γ -ray background.

Fig. 2.

Measurements of radial profiles for a line source in water. (A) is for an F-18 source while (B) is for an I-124 source. Each radial bin corresponds to 2.09-mm. Note the non-uniform offset present in the prompt-delays profiles for both radio-isotopes due to scattered coincidences. In addition, a flat offset is also present in the prompt-delays profile of I-124 due to the true coincidence γ -ray background.

Fig. 3.

Normalized time profiles for the F-18 and I-124 measurements for a line source in air ((A) and (C)) and water ((B) and (D)). (C) and (D) are plotted on logarithmic scale to accentuate any differences. There is a small increase in the dispersion of the I-124 time profiles as seen in the logarithmic plots. Each time bin corresponds to 25 ps.

Fig. 4.

Representative transverse slices from TOF reconstructed images for different setups. (A) F-18 after a 5-minute scan, (B) I-124 after a 5-minute scan and offset correction for true coincidence γ -ray background, (C) I-124 after a 20-minute scan and offset correction for true coincidence γ -ray background, (D) I-124 after a 5-minute scan and no special correction for true coincidence γ -ray background, and, (E) I-124 after a 20-minute scan and no special correction for true coincidence γ -ray background. In (D) and (E) the non-uniformity in the image is due to the presence of true coincidence events arising between one annihilation photon and the coincident 602 keV cascade γ , leading to an over subtraction of the estimated scatter at the center of the cylinder .

Fig. 5. Representative transverse slices from TOF reconstructed images for I-124 lesion phantom with varying amount of background I-131 present. (A) 0-MBq (0-mCi) of I-131, (B) 370-MBq (10-mCi) of I-131, (C) 740-MBq (20-mCi) of I-131, and (D) 1110-MBq (30-mCi) of I-131. The total amount of I-124 present in the lesion phantom was 40.7-MBq (1.1-mCi).

TABLE 1

Measured CRC values using F-18 and I-124 lesion phantoms

Sphere diameter (cm)	CRC (%)		
	F-18	I-124 (offset correction)	I-124 (no offset correction)
3.0	70±2	60±4	118±9
1.5	68±2	45±6	86±11
1.0	30±1	19±2	40±4

TABLE 2

Measured CRC values using an I-124 lesion phantom scanned for 20 mins in the presence of varying amounts of Background I-131. the total amount of I-124 in the lesion phantom was 10.7-MBQ (1.1-MCi) with a 6:1 uptake ratio for the spheres with respect to the background I-124 activity concentration

Sphere diameter (cm)	0-MBq (0-mCi) of I-131	370-MBq (10-mCi) of I-131	740-MBq (20-mCi) of I-131	1110-MBq (30-mCi) of I-131
CRC (%)				
3.0	73	77	90	96
1.5	61	72	81	89
1.0	23	25	30	36
Relative pixel noise (%)				
	9.6	12.8	22	31