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Towards Quantification of Functional Breast Images Using Dedicated SPECT With Non-Traditional Acquisition Trajectories

Kristy L. Perez,

Medical Physics Program and Radiology Department, Duke University, Durham, NC 27710 USA
(kristy.perez@duke.edu).

Spencer J. Cutler,

Biomedical Engineering and Radiology Departments, Duke University, Durham, NC 27710 USA
and also with GE Healthcare, Hillsborough, NC 27278 USA.

Priti Madhav[Member, IEEE], and

Biomedical Engineering and Radiology Departments, Duke University, Durham, NC 27710 USA
and also with GE Healthcare, Hillsborough, NC 27278 USA.

Martin P. Tornai[Senior Member, IEEE]

Medical Physics Program and Radiology Department, Duke University, Durham, NC 27710 USA
and also with the Biomedical Engineering and Radiology Departments, Duke University, Durham,
NC 27710 USA (martin.tornai@duke.edu).

Abstract

Quantification of radiotracer uptake in breast lesions can provide valuable information to physicians in deciding patient care or determining treatment efficacy. Physical processes (e.g., scatter, attenuation), detector/collimator characteristics, sampling and acquisition trajectories, and reconstruction artifacts contribute to an incorrect measurement of absolute tracer activity and distribution. For these experiments, a cylinder with three syringes of varying radioactivity concentration, and a fillable 800 mL breast with two lesion phantoms containing aqueous ^{99m}Tc pertechnetate were imaged using the SPECT sub-system of the dual-modality SPECT-CT dedicated breast scanner. SPECT images were collected using a compact CZT camera with various 3D acquisitions including vertical axis of rotation, 30° tilted, and complex sinusoidal trajectories. Different energy windows around the photopeak were quantitatively compared, along with appropriate scatter energy windows, to determine the best quantification accuracy after attenuation and dual-window scatter correction. Measured activity concentrations in the reconstructed images for syringes with greater than 10 $\mu\text{Ci}/\text{mL}$ corresponded to within 10% of the actual dose calibrator measured activity concentration for $\pm 4\%$ and $\pm 8\%$ photopeak energy windows. The same energy windows yielded lesion quantification results within 10% in the breast phantom as well. Results for the more complete complex sinusoidal trajectory are similar to the simple vertical axis acquisition, and additionally allows both anterior chest wall sampling, no image distortion, and reasonably accurate quantification.

Index Terms

Image processing; nuclear imaging; single photon emission computed tomography

I. Introduction

Quantification of absolute *in vivo* radiotracer uptake is not common for routine SPECT imaging. Generally, nuclear medicine physicians rely on differences in relative radioactive tracer uptake patterns to analyze molecular SPECT images. Because radiotracers can be

designed to label metabolic processes, receptors, and the like, their absolute *in vivo* quantification of concentration could be a valuable diagnostic tool to differentiate between benign and malignant tissue. Particularly in breast cancer patients, distinguishing absolute *in vivo* quantification via non-invasive molecular imaging could affect a patient's treatment plan.

However, quantification of the radioactive uptake in the images requires careful application of data corrections. Corrections for system factors, such as the detector, collimator and geometric efficiencies, must be incorporated.

Additionally, there are a variety of physical factors that can affect the accuracy of quantification in SPECT imaging [1]. These include photon attenuation and scatter which have both been shown to degrade images and hence, various methods have been proposed to correct for these. Attenuation correction relies on obtaining a spatial distribution of attenuation coefficients to model the imaged object, often derived from computed tomography (CT) data, and compensates for non-uniform attenuation.

There have been many scatter correction techniques derived over time for SPECT imaging [2]–[4]. The dual window (DW) method uses an energy window abutting the photopeak, but lower in energy, to estimate the scatter fraction in the photopeak [3].

Judiciously choosing narrow energy windows within the photopeak can minimize the scatter contributions and achieve better contrast in an image [5]. However, it is not clear if an energy window incorporating the lower energy tail of the photopeak would achieve better quantitative accuracy due to an increase in total counts contributing to the image.

Other groups have investigated quantitative breast imaging using planar single photon imaging techniques [6], [7] and have shown that quantification of size and uptake has implications for staging disease [6]. Due to the overlapping structures in planar imaging, it becomes difficult to correctly quantify smaller volumes. SPECT can localize the disease in 3D space and quantify its overlapping absolute radiotracer uptake without added activity from “normal” background tissue [1]. Because our dedicated SPECT system has a fully 3D range of motion, we are able to sample into the chest wall and axilla of the breast [8].

In this study, quantification is implemented and evaluated on our dedicated SPECT to test the system's linearity in measuring radiotracer uptake. In particular, the effect of data acquired with non-traditional, non-circular trajectories as well as the effect of energy window width on absolute quantification of SPECT data are characterized.

II. Methods

A. Gamma Camera and Data Acquisition

The SPECT sub-system of the hybrid imaging device consists of a Cadmium-Zinc-Telluride (CZT) *LumaGEM 3200STM* gamma camera (*Gamma Medica Inc.*, Northridge, CA) that has 2.5×2.5 mm pixels with a 6.7% intrinsic FWHM at 140 keV and a sensitivity of 37.9 cps/MBq (Fig. 1). The lead parallel hole collimator has 1.22 mm flat-to-flat hexagonal holes with 0.2 mm septa and is 25.4 mm in height. The camera system is attached to precision positioning motors to permit movement in 3D to contour the breast surface by moving in and out, up and down and around the center of rotation. Previous work has defined a set of trajectories which maximize the volume of the object imaged [8].

B. Model System in Reconstruction Code

A flow chart outlining the steps for quantifying emission data with this system is shown in Fig. 2. Image reconstruction was performed using a ray-driven, iterative ordered-subsets expectation maximization (OSEM) reconstruction code [9]. Collimator, geometry and detection efficiencies of the SPECT camera, radiopharmaceutical half-life, and attenuation and scatter correction maps were included in the reconstruction algorithm. In the reconstruction code, efficiencies and the half-life are constants input into the code, while the attenuation map is treated as a priori information. Scatter is treated as a constant to reduce reconstruction time [9]. The code employs the DW method [3] to estimate the scatter.

The reconstruction code has a scaling factor that increases or decreases the output, which must be set for each energy window. To find the reconstruction scaling factors, two 0.7 mL balloons (*Harvard Apparatus*, Holliston, MA) filled with 125.5 and 115.5 μCi of $^{99\text{m}}\text{Tc}$ pertechnetate in 0.5 and 0.4 mL were suspended at and away from the center of rotation, respectively, in a 12.5 cm diameter cylinder and were imaged with a simple circular, vertical axis of rotation (VAOR) acquisition trajectory in air. Different coefficients were obtained for each energy window used. All reconstructed slices containing the point sources were summed and a region of interest (ROI) was drawn encompassing all counts in the image. Equation 1 was used to determine the total activity in the reconstructed image.

$$A_{\text{tot}} = \sum_i \left[\left(\frac{A}{V} \right)_i V_i \right] \quad (1)$$

where A_{tot} is the total activity in the cumulative volume of interest, i is the voxel number, A/V is the reconstructed voxel value and V is the reconstructed voxel volume. The reconstruction scaling factor was determined by the ratio of the measured activity in the image to dose calibrator measured activity for each energy window.

C. Geometric & Anthropomorphic Phantoms

1. *Linearity and Calibration:* Three 30 mL syringes (2.1 cm inner diameter) were filled with ~ 10 mL of low concentrations of aqueous $^{99\text{m}}\text{Tc}$ pertechnetate, approximating clinical count rates [10], [11]. The syringes were placed in a 12.5 cm diameter cylinder (Fig. 3) and imaged in air, in water, and then in water plus background activity. The initial radioactive concentrations, measured with a calibrated dose calibrator (CRC-30BC, *Capintec, Inc.*, Ramsey, NJ), are given in Table I.
2. *Breast & Lesion Phantoms:* An 800 mL anthropomorphic breast phantom containing a 2.3 mL acrylic-walled sphere (*Radiological Service Devices Inc.*, Newport Beach, CA) and a 1.6 mL thin-walled balloon (*Harvard Apparatus*, Holliston, MA) were filled with aqueous $^{99\text{m}}\text{Tc}$ pertechnetate (Fig. 4). Table II gives the initial radioactive concentrations, measured with a dose calibrator, of the spheres and background.

D. Data Acquisition

For these data, 128 projection images collected over 360° with vertical axis of rotation (VAOR), 30° tilted parallel beam (TPB) and 3-lobed projected sinusoidal (PROJSINE) ranging from 15° to 45° polar tilt (Fig. 5) were compared for quantification accuracy. All data was collected in list mode to be resampled.

E. List Mode Resampling

Listmode data of the total counts acquired for all energies were re-sampled to produce projection files for symmetric $\pm 4\%$ (8%), $\pm 8\%$ (16%) and asymmetric $-16/+8\%$ (24%) photopeak energy windows. Appropriate uniformity corrections were applied to each energy window, derived from the original flood source list mode data (Fig. 6).

F. Attenuation and Scatter Corrections

A uniform emission volume to correct for signal loss due to photon attenuation was assumed. A map of the attenuation coefficients was defined by reconstructing the data to the first iteration, thresholding the image to obtain the mask of the object, and assigning each pixel a constant attenuation coefficient for water, 0.1545 cm^{-1} [12].

For scatter correction, the DW method estimates the percentage of scatter in the photopeak with a lower energy scatter window scaled by a proportionality constant, k [3]. Thus for a photopeak image, $f_1(x, y)$, and a scatter image, $f_2(x, y)$, then the equation for the corrected image is:

$$f(x, y) = f_1(x, y) - k f_2(x, y). \quad (2)$$

According to the DW method, the scatter window was centered around an energy that produced a window abutting and twice the size of the photopeak window (Fig. 6, Table III). The data acquisition system utilizes an electronic cutoff, and it does not record counts with energies below that limit, explaining why the Compton continuum does not plateau to 0 keV. Because k is a proportionality constant, it is expected that the trends in k are correct, but if the electronic cutoff is changed then k will scale appropriately. Following the methods outlined in [3], we determined that the DW method could be applied to our system and that the k value does not change with trajectory. For the energy windows used in this study, we empirically determined k for this setup iteratively using syringe 3 from the VAOR data collected in water.

G. Reconstruction

The original data collected in water and water plus background activity were reconstructed a second time implementing the attenuation map (derived from the first reconstruction) and scatter correction data. The data collected in air were reconstructed a second time implementing only the attenuation map. To determine the iteration at which the measured activity concentrations in the reconstructions approached convergence, the data set of syringes in air was reconstructed with 8 subsets up to 100 iterations with the OSEM reconstruction code. The mean value (units of microcuries per milliliter) in a ROI in the reconstructed image was obtained every 5 iterations to determine the iteration where the mean value levels off. A reconstruction grid size of $150 \times 150 \times 150$ was used. The isotropic voxel size was selected to be the same as the detector pixel size, with 2.5 mm on each side.

H. Data Analysis

Three sagittal slices of the reconstructed images were summed and ROIs were drawn in the syringe/lesions and backgrounds. ROIs for the syringes were completely within and not close to the edges of the syringes to avoid partial volume edge effects (1.5 cm wide by 1.75 cm long within the three central planes of the syringe). ROIs for the lesions (0.75 cm in diameter measured over the three central planes of the each lesion) were contained within, though closer to, the edges due to the smaller physical size of the lesions. The mean, decay corrected, reconstructed image activity concentration was determined and compared with the dose calibrator measured activity concentration. The percent difference was calculated to

determine the accuracy of the reconstruction process. All quantitative values were compared between acquisition trajectories and energy windows.

III. Results & Discussion

A. Convergence of Activity Concentration Value

The plot in Fig. 7 shows that for 8 subsets, the 20th iteration will provide information that is near convergence for each trajectory. It is known that this measurement is sensitive to ROI size, but the 20th did provided consistent results for each trajectory.

B. Linearity

The reconstructed data shows evident differences in contrast between the variety of activity concentrations (Fig. 8). The line profiles in Fig. 8 show that there is some variability within the syringes, which indicates the need for large ROIs within the syringe. The decay corrected, mean activity concentration is generally biased low, but within 15% of “known” values determined from dose calibrator measurements, which were considered the gold standard (Fig. 9). The results indicate that the accuracy has some dependence on activity concentration, energy window and acquisition trajectory. For lower activity concentrations, the accuracy is worse than for greater amounts of activity (Fig. 9). For example, in the $\pm 4\%$ and $\pm 8\%$ energy window when there is background activity present, the image measured activity concentration in syringe 1 is within 10 to 20% of the dose calibrator value compared to in syringe 3 where it is within 5 to 10%. Accuracy is generally within 10% when the activity concentrations are $> 3 \mu\text{Ci/mL}$ (syringes 2 and 3) in water and water plus background.

The images with syringes in air background have only attenuation correction, but the quantified values are consistently lower than the known activity concentration, except for syringes 2 and 3 collected with a PROJSINE trajectory and $-16/+8\%$ energy window. There seems to be a general trend for underestimating the activity concentration pointing to a need for better attenuation correction method. Additionally, it is unexpected that when the syringes are in an aqueous radioactive background all the quantification is biased low. Further investigation into improved attenuation correction with CT data is underway.

Not unexpectedly, the data with air background and no scatter correction shows the $-16/+8\%$ has the highest reported activity concentration for each syringe and is generally the most accurate. Furthermore, the estimation of activity in water was close to the true value; we expected that there would be a slight overestimation in the presence of background medium, but the opposite effect was observed. The $\pm 4\%$ and $\pm 8\%$ energy windows produce similar accuracies (Fig. 9).

For imaging this symmetric object, each acquisition trajectory produces reasonable results.

C. Breast & Lesion

VAOR and PROJSINE spherical lesion data are generally accurate to within 15% and 10%, respectively; while TPB data have consistently greater than 30% deviation from the actual dose calibrator measured concentration (Fig. 10). Because the TPB data is insufficiently sampled, elongation artifacts in the reconstructed images distort the shape of the breast and lesions, non-uniformly dispersing the activity into an incompletely sampled image volume. Additionally, there is an attenuation map inconsistency given the elongation artifact and the known, smoothly varying, true phantom distribution. This is caused by the incomplete sampling of the object. Therefore, the total activity is dispersed over the larger volume, and consequently the activity concentration reported in each voxel decreases, as evident in the

images set to the same global maximum in Fig. 11. From the present study, the cause for the overestimation of the background activity in the VAOR image is unknown. The background region in the cylinder data is also over estimated for the VAOR, TPB and PROJSINE trajectories by 18%, 23% and 16%, respectively. Further investigation into diffuse background regions is warranted.

The energy window data have generally similar accuracies, each within 15% of the known activity concentration.

IV. Conclusions

A method to quantify the activity concentration of regions of interest in data acquired with our unique dedicated SPECT system has been implemented. The average activity concentration reported in hot-volume regions corresponds to within 15% of the actual measured activity concentrations. For activity concentrations greater than 10 $\mu\text{Ci/mL}$, the measured activity concentration in large hot spots matched to within 10% of the dose calibrator value in the $\pm 4\%$ and $\pm 8\%$ energy window.

Larger errors persist in diffuse background regions, most likely due to the much higher noise characteristics (variability) in the background, especially in irregularly shaped objects such as the breast phantom. Very low activity concentrations are more difficult to accurately quantify, most likely due to the higher variability associated with fewer counts. The accuracy of quantifying low activity concentration regions by utilizing more of the photopeak events (increasing the counting statistics) was investigated. However, the largest energy window in most cases produced the least accurate quantification. This result could be due to the scatter correction method used, where counts were subtracted in proportion to a lower scatter energy window. For this very large energy window, the scatter proportionality constant was large, thus diminishing the otherwise perceived benefits of the large window. These results indicate that this energy window should not be selected for quantitative imaging.

In the cylinder and syringe data presented, the ability to quantify the activity concentration seems to be independent of acquisition trajectory. However when imaging the irregularly shaped breast, the TPB trajectory yielded the well-known, distorted data that was not as accurately quantified as the VAOR and PROJSINE trajectories. Less extreme tilt for the TPB trajectory has indeed shown less distortion [14], which could lead to better quantification with that trajectory. Moreover, use of attenuation or scatter correction in such a clearly distorted reconstructed volume doesn't make sense. The data indicates that either the VAOR or PROJSINE trajectories would provide adequate data for quantification. VAOR is the only complete sampling trajectory used here, but also impractical, given the nature of pendant breast imaging: the gamma camera with parallel-hole collimator cannot image the chest wall with VAOR. Thus, the result that PROJSINE yields nearly the same result as for VAOR indicates that it is the preferred trajectory in clinical imaging: it allows both anterior chest wall sampling with a more complete acquisition trajectory, no image distortion, and reasonably accurate quantification. Future steps should determine the accuracy of quantifying hot spots in the chest wall region with the PROJSINE trajectory.

Additional future steps to improve results would include using the CT reconstructed data set to calculate the spatial distribution map of attenuation coefficients as well as to define the ROI for better placement, and correcting for partial volume effects. To analyze the mean activity concentration in the syringe, relatively large ROIs were used and placed well within the boundaries of the syringe, minimizing partial volume effects. However, for smaller and/or irregularly shaped hot spots (lesions), ROI placement has a large effect on the measured

mean activity concentration. Thus, having a CT selected ROI would give more confidence in the measurement.

In conclusion, the $\pm 8\%$ energy window had slightly better accuracy than the $\pm 4\%$ energy window, but either could be considered for future experiments and patient imaging. VAOR or PROJSINE trajectories should be used for the best quantitative results.

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References

1. Rosenthal MS, Cullom J, Hawkins W, Moore SC, Tsui BM, Yester M. Quantitative SPECT imaging: A review and recommendations by the focus committee of the society of nuclear medicine computer and instrumentation council. *J. Nucl. Med.* 1995 Aug.;vol. 36:1489–1513. [PubMed: 7629599]
2. Axelsson B, Msaki P, Israelsson A. Subtraction of Compton-scattered photons in single-photon emission computerized tomography. *J. Nucl. Med.* 1984 Apr.;vol. 25:490–494. [PubMed: 6400024]
3. Jaszczak RJ, Greer KL, Floyd CE Jr, Harris CC, Coleman RE. Improved SPECT quantification using compensation for scattered photons. *J. Nucl. Med.* 1984 Aug.;vol. 25:893–900. [PubMed: 6611390]
4. Kadrmas DJ, Frey EC, Karimi SS, Tsui BM. Fast implementations of reconstruction-based scatter compensation in fully 3D SPECT image reconstruction. *Phys. Med. Biol.* 1998 Apr.;vol. 43:857–873. [PubMed: 9572510]
5. Cutler SJ, Brzymialkiewicz CN, Tornai MP. Investigating the effects of energy resolution in dedicated emission mammothomography. *Proc. IEEE Nuclear Science Symp. Conf. Rec.* 2005:2537–2541.
6. Hruska CB, O'Connor MK. Quantification of lesion size, depth, and uptake using a dual-head molecular breast imaging system. *Med. Phys.* 2008 Apr.;vol. 35:1365–1376.
7. Williams MB, More MJ, Narayanan D, Majewski S, Welch B, Wojcik R, Kieper D. Phantom study of radiotracer concentration quantification in breast scintigraphy. *IEEE Trans. Nucl. Sci.* 2003 Jun.; vol. 50(no. 3):433–438.
8. Archer CN, Tornai MP, Bowsher JE, Metzler SD, Pieper BC, Jaszczak RJ. Implementation and initial characterization of acquisition orbits with a dedicated emission mammothomograph. *IEEE Trans. Nucl. Sci.* 2003 Jun; vol. 50(no. 3):413–413.
9. Bowsher JE, Johnson VE, Turkington TG, Jaszczak RJ, Floyd CR, Coleman RE. Bayesian reconstruction and use of anatomical a priori information for emission tomography. *IEEE Trans. Med. Imag.* 1996; vol. 15:673–686.
10. Maublant J, de Latour M, Mestas D, Clemenson A, Charrier S, Feillel V, Le Bouedec G, Kaufmann P, Dauplat J, Veyre A. Technetium-99m-sestamibi uptake in breast tumor and associated lymph nodes. *J. Nucl. Med.* 1996; vol. 37:922. [PubMed: 8683312]
11. Wackers FJ, et al. Technetium-99m hexakis 2-methoxyisobutyl isonitrile: Human biodistribution, dosimetry, safety, and preliminary comparison to thallium-201 for myocardial perfusion imaging. *J. Nucl. Med.* 1989; vol. 30:301. [PubMed: 2525610]
12. Hubbell, JH.; Seltzer, SM. NIOSA Technology, Ed. National Institute of Standards and Technology; 1996. Tables of x-ray mass attenuation coefficients and mass energy-absorption coefficients from 1 keV to 20 MeV for elements $Z = 1$ to 92 and 48 additional substances of dosimetric interest.

13. Brzymialkiewicz, CN. Ph.D. dissertation. Durham, NC: Dept. Biomed. Eng., Duke Univ.; 2005. Development and evaluation of a dedicated emission mammotomography system.
14. Perez KL, Cutler SJ, Madhav P, Tornai MP. Characterizing the contribution of cardiac and hepatic uptake in dedicated breast SPECT using tilted trajectories. *Phys. Med. Biol.* 2010 Aug.;vol. 55:4721–4734. [PubMed: 20671354]

Fig. 1.
Hybrid SPECT-CT breast imaging device.

Fig. 2.
Flowchart of the steps for emission data quantification.

Fig. 3.
Photograph of syringes with varying concentrations of radioactivity in a cylinder with an aqueous uniform radioactive background.

Fig. 4. Photograph of the acrylic (left) and balloon (right) spheres in the anthropomorphic breast phantom. The black arrow indicates the water level at the time of imaging, which is not pictured to avoid distortion due to refraction.

Fig. 5.
3D trajectories of the gamma camera used to acquire data.

Fig. 6.

Example energy spectra showing the energy windows used in the list mode resampling to obtain $\pm 8\%$ and $-16 + 8\%$ photopeaks (solid black and gray lines, respectively) and corresponding scatter windows (dotted black and gray upper and lower threshold lines, respectively) data for reconstruction.

Fig. 7. Plot of mean activity concentration in an ROI as a function of iteration number. Iteration 20 was determined to be a good compromise for stopping iterations for each of the trajectories.

Fig. 8.

Average of seven summed reconstructed slices of syringes in air (TOP, LEFT), water (MIDDLE, LEFT) and aqueous radioactivity background (BOTTOM, LEFT) acquired with a VAOR trajectory and a $\pm 4\%$ photopeak energy window; images from other trajectories look similar. Three pixel line profiles (RIGHT COLUMN) through each syringe (S1 = 3.6, S2 = 11, S3 = 21 $\mu\text{Ci/mL}$) for each acquisition.

Fig. 9.

Bar charts of dose calibrator (black lines) and decay corrected image measured activity concentrations for 8% (dark grey on left), 16% (light grey in middle) and 24% (medium grey on right) energy windows for syringes 1 (TOP), 2 (MIDDLE), and 3 (BOTTOM) and for air (LEFT), water (CENTER) and aqueous radioactive (RIGHT) backgrounds. Note that the air data is not scatter corrected.

Fig. 10.

Bar charts of dose calibrator (black lines) and decay corrected image measured activity concentration for the acrylic (TOP) and balloon lesions (CENTER) and uniform breast background (BOTTOM).

Fig. 11.

Three summed reconstructed slices smoothed with a Gaussian kernel in imageJ. VOAR (TOP), TPB (MIDDLE) and PROJSINE (BOTTOM) images are set to a global maximum and contain acrylic (LEFT) and thin-walled (RIGHT) radioactive lesion phantoms.

TABLE I

Initial Radioactivity and Activity Concentration in Syringes and Cylinder

	Activity (μCi)	Volume (mL)	Activity Cone ($\mu\text{Ci/mL}$)
Syringe 1	34.7	9.7	3.6
Syringe 2	113	10	11
Syringe 3	227	11	21
Background	387	900	0.36

TABLE II

Initial Radioactivity and Activity Concentration of the Spheres and Breast Phantoms

	Activity (μCi)	Volume (mL)	Activity Cone ($\mu\text{Ci/mL}$)
Acrylic	28.4	2.3	12
Balloon	19.8	1.6	12
Background	223	765	0.29

TABLE III

Scatter Energy Windows and Scatter Fractions for Each Photopeak Energy Window

Photopeak Window	Photopeak Range (keV)	Scatter Window Range (keV)	Proportionality Constant (k)
$\pm 4\%$	134 – 147	113 – 133	0.16
$\pm 8\%$	129 – 151	92 – 127	0.26
$-16+8\%$	118 – 151	71 – 116	0.92