Published in final edited form as:

IEEE Nucl Sci Symp Conf Rec (1997). 2010 ; 2010: 2319–2324. doi:10.1109/NSSMIC.2010.5874198.

Is SPECT or CT Based Attenuation Correction More Quantitatively Accurate for Dedicated Breast SPECT Acquired with Non-Traditional Trajectories?

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Abstract

Attenuation correction is necessary for SPECT quantification. There are a variety of methods to create attenuation maps. For dedicated breast SPECT imaging, it is unclear if either SPECT- or CT-based attenuation map would provide the most accurate quantification and whether or not segmenting the different tissue types will have an effect on the quantification. For these experiments, ^{99m}Tc diluted in methanol and water was filled into geometric and anthropomorphic breast phantoms and was imaged with a dedicated dual-modality SPECT-CT scanner. SPECT images were collected using a compact CZT camera with various 3D acquisitions including vertical and 30° tilted parallel beam, and complex sinusoidal trajectories. CT images were acquired using a quasi-monochromatic x-ray source and CsI(Tl) flat panel digital detector in a half-cone beam geometry. Measured scatter correction for SPECT and CT were implemented. To compare photon attenuation correction in the reconstructed SPECT images, various volumetric attenuation matrices were derived from 1) uniform SPECT, 2) uniform CT, and 3) segmented CT, populated with different attenuation coefficient values. Comparisons between attenuation masks using phantoms consisting of materials with different attenuation values show that at 140 keV the differences in the attenuation between materials do not affect the quantification as much as the size and alignment of the attenuation map. The CT-based attenuation maps give quantitative

values 30% below the actual value, but are consistent. While the SPECT-based attenuation maps can provide within 10% accurate quantitative values, but are less consistent.

I. Introduction

A variety of physical factors affect the quantification of single photon emission computed tomography (SPECT) images [1]. Image artifacts arise from physical processes and sampling, especially with systems capable of 3D trajectories, which can yield an incorrect absolute activity of the tracer. Photon attenuation in the body impairs image quantification by reducing the total number of counts detected by the SPECT camera. Attenuation correction relies on obtaining a spatial distribution of attenuation coefficients to model the imaged object and compensates for non-uniform attenuation. For some imaging tasks, assuming a constant attenuation is sufficient [2]. However, this distribution can also be derived from computed tomography (CT) data.

Using CT-based attenuation correction must account for a variety of factors pertaining to the differences in the SPECT and CT modalities, such as different resolutions, imaging (and attenuation) energies, and, for our system in particular, different imaged volumes and orientations. The differences in resolution and alignment have been shown to be problematic when using a CT image to correct SPECT studies. Misregistration of 2 to 3 cm has been shown to cause a 20 to 35% change in radiotracer distribution [3]. Another study found that a 7 mm misalignment caused a 15% change in apparent radiotracer uptake [4]. These errors can be introduced by the registration software or by the user variability interacting with registration software [5]. Using the SPECT reconstruction does not introduce these errors, and may be an appropriate method considering the nearly uniform attenuation of varying breast tissues at 140 keV.

In this study, attenuation maps based on reconstructed SPECT and CT data are compared. CT-based attenuation maps were created both by linearly scaling the attenuation coefficients measured at 36 keV to 140 keV as well as filling in the known attenuation coefficient at 140 keV. The CT-based attenuation maps created applied one correction/value to the data as well as segmented the data and then applied the corrections/values for water and methanol separately. The effect of the different SPECT acquisition trajectories was also observed.

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 3200S*TM gamma camera (*Gamma Medica Inc.*, Northridge, CA) that has 2.3×2.3 mm crystals on a 2.5 mm pitch 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 [6].

The CT sub-system has a rotating tungsten target x-ray source (*Rad-94, Varian Medical Systems*, Salt Lake City, UT), with a 0.4 focal size and 14° anode angle, and a 20×25 cm² CsI(Tl)-based amorphous silicon digital x-ray detector (*Paxscan 2520, Varian Medical Systems*, Salt Lake City, UT) with a grid size of 1920×1536 pixels on a 127 μm pitch (Fig. 1). The x-ray source and detector are attached to U-shaped metal plate securing the two at a fixed tilt and are adjoined to a common azimuthal rotation stage (model RV350CCHL, *Newport Corp.*, Irvine, CA). The collimator attached to the x-ray source holds a cerium 100th attenuating value layer (0.0508cm) filter ($Z=58$, $\rho=6.77\text{g/cm}^3$, $K\text{-edge}=40.4\text{keV}$, *Santoku America, Inc.*, Tolleson, AZ), which yields a mean beam energy of ~36 keV and FWHM of 15% [7].

B. Geometric & Anthropomorphic Phantoms

1) Cylindrical Phantom—The cylindrical phantom consisted of 4 syringes filled with ^{99m}Tc-pertechnetate diluted in 10 mL of methanol or water. The initial radioactive concentrations, measured with a calibrated dose calibrator (CRC-30BC, *Capintec, Inc.*, Ramsey, NJ), are given in Table I. The syringes were placed in a 12.5 cm diameter cylinder (Fig. 2) filled with aqueous ^{99m}Tc-pertechnetate. Additionally, four 6 mm diameter nylon spheres that had been soaked in 1 mCi of ^{99m}Tc-pertechnetate for ~1 hour were taped to the outside of the cylinder. Due to the opposite charges of the pertechnetate and nylon, the radioactive molecule collects on the surface of the sphere and is used as a fiducial marker to register SPECT and CT images.

2) Breast & Lesion Phantoms—A 700 mL anthropomorphic breast phantom contained two 5.4 mL acrylic-walled spherical lesions (*Radiological Service Devices Inc.*, Newport Beach, CA), one filled with ^{99m}Tc pertechnetate diluted into methanol and the other with ^{99m}Tc pertechnetate diluted into water (Fig. 3). Table II gives the initial radioactive concentrations, measured with a dose calibrator, of the spheres and background. Additionally, four 6 mm diameter, ^{99m}Tc-pertechnetate soaked, nylon spheres were taped to the outside of the breast phantom for image registration.

C. Data Acquisition

For this data, 128 projection images collected over 360° with vertical axis of rotation (VAOR), 30° tilted parallel beam (TPB) and sinusoidal wave projected onto a hemisphere ranging from 15° to 45° polar tilt (PROJSINE) trajectories (Fig. 4) were compared for quantification accuracy. The data was collected with a ±4% photopeak energy window centered about 140 keV.

Additionally, 240 x-ray projection images acquired with 60 kVp and 1.25 mAs were collected both without and with a lead-bead scatter grid in place, such that the projections could be scatter corrected prior to reconstruction, resulting in nearly absolute attenuation coefficients.

D. Creating Attenuation Maps & Reconstructing Data

SPECT image reconstruction was performed using a ray-driven, iterative ordered-subsets expectation maximization (OSEM) reconstruction code [8]. CT image reconstruction used an OSC iterative reconstruction code.

For SPECT based attenuation correction, 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} (Fig. 5, TOP). For CT based attenuation correction, scatter correction, reconstructed CT images were 1) filled with a constant attenuation coefficient of water, 0.1545 cm^{-1} , 2) segmented and filled with attenuation coefficients of water, 0.1545 cm^{-1} , and methanol, 0.1234 cm^{-1} , 3) water attenuation coefficient linearly scaled from 36 keV to 140 keV and 4) segmented water and methanol attenuation coefficients linearly scaled from 36 keV to 140 keV (Fig. 5).

The original data was reconstructed a second time implementing with each of the five attenuation maps as well as the collimator, geometry and detection efficiencies of the SPECT camera, radiopharmaceutical half-life, and scatter correction maps were included in the reconstruction algorithm as previously described [9]. The data was reconstructed to the 20th iteration, near convergence. 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. Thus, gray scale values of the reconstructed images are output in absolute $\mu\text{Ci/mL}$ units.

E. Data Analysis

Volumes of interest (VOIs) for the syringes were completely within and not close to the edges of the syringes to avoid partial volume edge effects. 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.

III. Results

A. Cylinder Phantom

There were no obvious qualitative differences in the images reconstructed with the different attenuation maps. However, it was noted that if CT-based attenuation maps are not correctly aligned with respect to where the SPECT reconstruction code places the image within the matrix, artifacts and quantitative inaccuracies will result. The reconstructed images were very similar (Fig. 6).

Quantitatively, the SPECT-based attenuation correction was more accurate than the CT-based attenuation correction (Fig. 7). However, the CT-based attenuation correction produced more consistent results of approximately 73% of the known value.

B. Breast & Lesion Phantoms

The reconstructed images for each trajectory with each attenuation map are very similar (Fig. 8). The SPECT-based attenuation corrected mean activity concentration has good agreement with “known” values, while CT-based attenuation correction is less accurate, but consistent as seen in the cylinders (Fig. 9).

IV. Discussion

The SPECT-based attenuation map has better accuracy than the CT-based attenuation correction, but the CT produces more precise results. However, one caveat is the quantification method was first developed with the SPECT-based attenuation map and thus maybe biased towards it. Nonetheless, the CT-based attenuation correction quantification data implies that accounting for the difference in attenuation coefficients at 140 keV of water (0.1545 cm^{-1}) and methanol (0.1234 cm^{-1}) does not make a difference in the quantitative accuracy. The CT-based attenuation maps were slightly more accurate when using the water only attenuation coefficient even when quantifying the activity concentration of methanol syringe/lesion. Therefore, it seems feasible to either use a uniform SPECT-based attenuation map or scale the CT data with a constant linear scaling term in the future.

The CT-based attenuation maps are consistently between 20 to 35% lower than the known activity concentration, similar to previous results seen in our lab, where the known absolute volume was used to quantify SPECT images [10]. Since the CT-based attenuation correction appears more consistently reproducible, after scaling adjustments accounting for volume mismatch, it may be more clinically useful. Alternatively, if resolution recovery was employed in the SPECT reconstruction, perhaps the reduced blur and more accurate volumes in the SPECT images coupled with the CT images would produce more accurate and reproducible quantification. This topic will be explored in future research.

It should be noted that VOI placement can affect the apparent accuracy of quantification. For this data set, large VOIs completely within the object of interest were selected to minimize partial volume sampling. However, changing the VOI placement could report different mean image values, but not different trends.

Images were registered in AMIDE where the uncertainty of registration is on the order of 2 to 3mm/point, based on the potential for mis-identifying the center of a fiducial marker. Given that it has been shown in literature that a 7mm shift can cause up to 15% inaccuracy in radioactivity distribution [4], this mis-alignment could reduce the accuracy of quantification when the CT-based attenuation maps are used. Additional alignment errors could be introduced when writing out rotated images in AMIDE because the software writes the images out to a bigger matrix that has to be cropped, introducing potential manual alignment errors. Furthermore, when AMIDE interpolates between voxels to rotate and align the attenuation map, new voxel values may have adjusted attenuation coefficient values, especially at the edge of the object or at interfaces between segmented mediums (methanol and water). Further investigation into image registration software should be done in the future to reduce these issues.

V. 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 SPECT-based attenuation map currently provides the best quantification, but the CT-based attenuation correction could be improved and provide more consistent results.

MPT is an inventor of this dedicated SPECT imaging technology, and is named as an inventor on the patent for this technology assigned to Duke. If this technology becomes commercially successful, MPT and Duke could benefit financially.

Acknowledgments

This work has been funded by the National Cancer Institute of the National Institutes of Health (R01-CA096821, T32-EB001040, and T32EB007185) and the Department of Defense Breast Cancer Research Program (W81XWH-08-1-0192).

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Fig. 1. Hybrid SPECT-CT breast imaging device. Orange and yellow arrows indicate the directions of movement of the SPECT camera (center). X-ray source (right) and detector (left) orbit the center-of-rotation at a fixed tilt.

Fig. 2.

Photograph of syringes with varying concentrations of radioactivity in a cylinder with an aqueous uniform radioactive background. Green syringes are filled with methanol, and purple syringes are filled with water. Medical tape attached to outside of the container secure fiducial markers used to register SPECT and CT images.

Fig. 3. Photograph of the water-filled (purple) and methanol-filled (green) spheres in the anthropomorphic breast phantom filled with 700 mL of water. Medical tape attached to outside of the breast secures fiducial markers used to register SPECT and CT images.

Fig. 4.
3D trajectories (yellow path) of the gamma camera (blue box) used to acquire data of the cylinder and breast phantoms.

Fig. 5.

Maps used to attenuation correct SPECT data for quantification. (TOP to BOTTOM) SPECT-based attenuation map and CT-based attenuation maps with 1) constant value of water, 2) segmented, constant values of water and methanol, 3) water attenuation coefficient linearly scaled from 36 to 140 keV and 4) segmented water and methanol attenuation coefficients linearly scaled from 36 to 140 keV.

Fig. 6. 20th iteration, Gaussian smoothed, 3 summed reconstructed slices of the cylinder and syringes in the (LEFT) coronal and (RIGHT) transverse planes. Images are SPECT-based attenuation corrected images collected with (TOP) VAOR, (MIDDLE) TPB and (BOTTOM) PROJSINE trajectories.

Fig. 7.

Bar charts of the ratio of image measured activity concentration to the dose calibrator known activity concentration of each syringe for each attenuation map and each SPECT acquisition trajectory. A ratio of one indicates the perfect image quantification.

Fig. 8.

Reconstructed, attenuation corrected SPECT images of the breast with methanol and water lesions collected with (TOP) VAOR, (MIDDLE) TPB and (BOTTOM) PROJSINE trajectories.

Fig. 9.

Bar charts of the ratio image measured activity concentration to the dose calibrator known activity concentration of each lesion for each attenuation map and each SPECT acquisition trajectory. A ratio of one indicates the best quantification.

Table I

Initial activity concentration in syringes and syringe-to-background ratio in cylinder.

	Activity Concentration (uCi/mL)	Syringe:Bkgd Ratio
Methanol High	33.6	8:1
Methanol Low	16.0	4:1
Water High	32.3	8:1
Water Low	16.9	4:1

Table II

Initial activity concentration of the lesions in the breast phantom and the lesion-to-background ratio.

	Activity Concentration (uCi/mL)	Lesion:Bkgd Ratio
Methanol	26.2	7:1
Water	27.0	7:1