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Design and characterization of a low profile NaI(Tl) gamma camera for dedicated molecular breast tomosynthesis

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Abstract

A new low profile gamma camera is being developed for use in a dual modality (x-ray transmission and gamma-ray emission) tomosynthesis system. Compared to the system's current gamma camera, the new camera has a larger field of view (~20×25 cm) to better match the system's x-ray detector (~23×29 cm), and is thinner (7.3 cm instead of 10.3 cm) permitting easier camera positioning near the top surface of the breast. It contains a pixelated NaI(Tl) array with a crystal pitch of 2.2 mm, which is optically coupled to a 4×5 array of Hamamatsu H8500C position sensitive photomultiplier tubes (PSPMTs). The manufacturer-provided connector board of each PSPMT was replaced with a custom designed board that a) reduces the 64 channel readout of the 8×8 electrode anode of the H8500C to 16 channels (8X and 8Y), b) performs gain non-uniformity correction, and c) reduces the height of the PSPMT-base assembly, 37.7 mm to 27.87 mm. The X and Y outputs of each module are connected in a lattice framework, and at two edges of this lattice, the X and Y outputs (32Y by 40X) are coupled to an amplifier/output board whose signals are fed via shielded ribbon cables to external ADCs. The camera uses parallel hole collimation. We describe the measured camera imaging performance, including intrinsic and extrinsic spatial resolution, detection sensitivity, uniformity of response, energy resolution for 140 keV gamma rays, and geometric linearity.

Keywords

nuclear medicine; gamma camera; breast imaging

1. INTRODUCTION

Over the last 20 years, research has demonstrated the value of compact high resolution gamma cameras for dedicated nuclear medicine breast imaging as a means of mitigating the lack of resolution and sensitivity of standard large cameras^{1–5}. Three main types of gamma

cameras are currently being implemented: scintillation cameras with position sensitive photomultiplier tubes (PSPMTs) for PET and SPECT, scintillation cameras using silicon photomultipliers (SiPMs) for PET, and solid state direct-conversion semiconductor cameras for PET and SPECT. PSPMT based gamma cameras for breast imaging typically contain a pixelated array of sodium iodide (NaI) or cesium iodide (CsI) crystals, in which confinement of the scintillation light to individual crystals allows for a smaller dead space at the chest wall edge of the detector compared to single-crystal designs^{2,6}. SiPMs are high gain integrated arrays of parallel avalanche diodes that operate in Geiger mode and are compact alternatives to bulkier PMTs. They have been successfully used in dedicated breast PET applications in which the 511 keV annihilation photons result in a large number of light photon from the crystal. However their larger electron multiplication gain variance and resulting poorer energy resolution compared to PMTs has limited their application in lower energy single gamma systems to date⁷⁻⁹. The most commonly used solid-state semiconductor is cadmium zinc telluride (CZT). CZT arrays exhibit good gamma ray detection efficiency and have been shown to have very good energy resolution^{5,10-12}, but suffer from low energy tailing due to hole trapping. CZT systems are currently available from a small number of sources at a great cost for low-defect samples.

The PSPMT-based scintillation camera described here was developed as a replacement for a gamma camera used in a dual modality tomosynthesis system (DMT). The DMT system consists of an x-ray transmission component and a gamma-ray emission component, each performing limited angle tomographic imaging and sharing a common breast immobilization system allowing for unambiguous correlation between radiotracer distribution and anatomic structures in the breast (Figure 1). The main design considerations for the new gamma camera were to a) increase the field of view (FOV) from 15 cm $\times$ 20 cm to 20 cm $\times$ 25 cm, in order to better match the 23 cm $\times$ 29 cm FOV of the DMT system's x-ray detector, b) improve system sensitivity and overall imaging performance, and c) decrease the camera thickness, allowing for better camera positioning and patient comfort.

2. MATERIALS AND METHODS

2.1 Camera Design

The low profile (LP) gamma camera contains a pixelated NaI(Tl) scintillator with a crystal size of 1.95 mm by 1.95 mm and 6 mm thick with a crystal pitch of 2.2 mm. The overall active area is 250.5 mm $\times$ 199.9 mm. The crystal array's 6.5 mm thick window is optically coupled to a 4 $\times$ 5 array of 5 cm $\times$ 5 cm Hamamatsu H8500C PSPMTs. In order to compensate for gain variations among the PSPMTs, reduce the number of data channels necessary for readout, and add signal amplification, the manufacturer-provided connector board of each PSPMT was replaced with customized gain matching X-Y conversion board (Figure 2a). Each customized board contains a gain-matching resistor matrix based on the manufacturer-provided gain map of each PSPMT. The board reduces the 64 channel readout of the 8 $\times$ 8 electrode anode array of each H8500C to 16 channels (8X and 8Y) by connecting the rows and columns of the PSPMT anodes. The custom boards reduce the total height of each PSPMT module from 37.7 mm to 27.8 mm (Figure 2b). The final total height of the PSPMTs, matrix/gain matching boards, and amplifier/summing board with connectors

comes to 37.8 mm, almost exactly the same as the original raw PSPMT height. This is a reduction of almost 2 cm (30%) from our previous PSPMT/electronics configuration.

The X and Y outputs of each board are connected in a lattice framework that is read out along two edges of the camera (40 X by 32 Y) and coupled to an amplifier/summing board. Each signal goes through a low-noise pre-amplification stage followed by a second amplification stage and then sent out of the detector head on shielded ribbon cables. All of 40 X signals are also summed and, after discrimination, used as detector trigger. The signals on the shielded ribbon cables are sent to MDU40-GI32 multichannel integrating ADCs (AiT Instruments, Newport News, VA), see (Figure 3a,b). The camera employs a parallel hole collimator with a hexagonal hole height of 18.4 mm, septal thickness of 0.26 mm, and face-to-face diameter of 1.9 mm (Figure 3c). Before characterization, the LP gamma camera went through a calibration process consisting of measuring the pedestals of each ADC channel, mapping the locations of the pixelated crystals in the raw images, and obtaining crystal-specific pulse height spectra to produce a gain correction lookup table. To determine the ADC pedestals, triggers were generated with the PSPMT high-voltage supply turned off. This was repeated multiple times to test pedestal stability. For crystal mapping a high statistics raw image was obtained using a Tc-99m source (Figure 4). Once the crystals were mapped, the energy calibration was completed. Characterization of the system's imaging properties was performed according to both NEMA NU-1 standards and commonly published gamma camera characterization methodologies^{1,2,13}.

2.2 Energy Resolution

Good energy resolution permits sensitive rejection of scattered gamma-rays. The energy resolution is dependent primarily on the light output of the scintillator array and the quantum efficiency and gain characteristics of the PSPMTs. To measure the energy resolution a Tc-99m point source at a distance of 125 cm from the camera face was used to flood the entire field of the gamma camera (with the collimator removed) for a sufficient period of time for an individual pulse height distribution to be obtained for each crystal. In each crystal's spectrum the 140 keV photopeak was identified and used to create a lookup table of relative gains. The combined energy spectrum was obtained by summing the individual crystal spectra after normalization by the relative gain factors so that the centers of the photopeaks fell in the same spectral bin (Figure 5a). This experiment was then repeated with a Co-57 and Am-241 source. A plot of peak energy versus bin number was created and fit linearly to find a calibration factor (keV per bin) (Figure 5b). The energy resolution is reported as the full width at half maximum (FWHM) of the photopeak in bins multiplied by the calibration factor then divided by the center photopeak energy of Tc-99m of 140.5 keV¹³.

2.3 Intrinsic Uniformity

Mapped gamma camera images are corrected for spatially dependent camera sensitivity non-uniformities using a high-statistics correction image obtained under conditions of uniform gamma ray fluence. To correct sensitivity non-uniformities, the raw image is divided on a pixel-by-pixel basis by the high statistics flood image, which has been normalized by dividing each pixel value by the average pixel value. The correction image is acquired with a

minimum of 10,000 counts per channel, in order to minimize the effect on the corrected image of added statistical noise. The NEMA protocol requires that the uniformity of both the useful field of view (UFOV) and the central field of view (CFOV) are reported. The UFOV for this gamma camera is 250.5 mm $\times$ 199.9 mm (the full active area of the crystal array). The CFOV is defined as the central 56.3% of the UFOV area (75% of each UFOV dimension), which in this case is 187.8 mm $\times$ 149.9 mm. To test the camera's intrinsic uniformity, the collimator was removed and a Tc-99m point source was placed at a distance of 125 cm from the camera.

For both the UFOV and CFOV both integral and differential non-uniformity were calculated and are reported in accordance to the NEMA protocol. To calculate the integral uniformity, the maximum and minimum pixels for each FOV are found. The integral non-uniformity is defined as the difference between the maximum and minimum pixel values, divided by the sum of the maximum and minimum pixels and it is reported as a percentage. Differential non-uniformity is calculated by finding the largest difference between any two pixels within a set of 5 contiguous pixels in a row or column. The differential uniformity is then defined as the difference divided by the sum of the maximum and minimum pixel values within the 5 contiguous pixels. This is done for both the X and Y directions for each column and the maximum value is reported for each direction¹³.

2.4 Intrinsic Geometric Linearity

The primary source of geometric non-linearity in detectors with tiled PSPMTs is biasing of the measured scintillator light centroid position due to the way that the scintillation light is shared between more than one PSPMT near PSPMT edges. Additional distortions are caused by truncation of the amplified charge pulse near the PSPMT perimeters. Figure 4 shows a high resolution raw image made with a uniform gamma ray illumination. The clustering of the detected events is caused by the light piping effect of the individual crystals. The clusters do not lie along straight row and column lines because of the light sharing and charge truncation effects. However, these distortions can be corrected by mapping the events of each cluster onto a regular matrix corresponding to the known locations of each crystal in the array, leaving only a few small non-linearities associated with any array imperfections (Figure 6). The residual geometric distortions following crystal mapping were measured with a 0.5 mm slit created using a lead mask. The slit was placed in the center of the UFOV and was positioned at a 45-degree angle². A Tc-99m point source was placed at a distance that permitted illumination of the entire slit. The spatial differential linearity was calculated by fitting a linear regression to the peak points of the data and the original data points were then subtracted from the fitted line.

2.5 Intrinsic Spatial Resolution

The lower limit on the intrinsic (no collimator) spatial resolution of gamma cameras with pixelated crystals gamma cameras is set by the crystal pitch, however, the resolution is degraded by gamma-ray scattering, which can deposit energy in more than one crystal. Resolution is also degraded by optical crosstalk between crystals and by the spatial resolution of the photomultipliers. To measure the intrinsic spatial resolution a 0.25 mm wide, 4 mm thick lead slit placed directly on the detector input surface was used. The

gamma camera was irradiated with a Tc-99m point source positioned at a distance of 125 cm. The slit was oriented at a slight angle with respect to the crystal array matrix so that the offset between the line source midline and the crystal centers varied along the length of the slit source. By subsampling profiles extracted along rows, and shifting them to align the centers of the slit image, the profiles could be summed to yield a spatially averaged line spread function (LSF)². The averaged LSF was fitted with a Gaussian and the FWHM and FWTM were reported¹³.

2.6 Extrinsic Spatial Resolution

Due to the collimator's geometric spatial resolution the extrinsic resolution (with collimator) of the gamma camera is lower than the intrinsic resolution. Because, unlike the intrinsic resolution, the collimator resolution varies with the source-to-collimator distance, the extrinsic spatial resolution was measured as a function of the source-to-collimator separation. In accordance with the NEMA protocol the resolution both with and without scatter were measured. For the resolution without scatter, a small diameter capillary tube (inside diameter of 0.9 mm) filled with a Tc-99m solution was moved away from the collimator in 1 cm steps for 10 cm. For each source-to-collimator distance the capillary tube was moved laterally at 2 mm steps for 10 mm. For each height, the FWHM and FWTM of each of the lateral positions were found via Gaussian fits and averaged to obtain the average spatial resolution. To obtain a measurement of spatial resolution including the effects of scatter a 10 cm block of acrylic was placed between the source and the collimator.

Only one source distance was measured for this resolution, but as before the source was moved laterally at 2 mm steps over 10 mm with the final resolution reported as an average of the FWHM of the lateral positions¹³.

2.7 System Sensitivity

The largest factors affecting the camera sensitivity are the geometric collimator sensitivity, the absorption efficiency of the crystal material, the open area fraction of the crystal array, and dead time created by the system's electronics and acquisition software. The sensitivity was measured according to the NEMA protocol. The bottom of a 138 mm diameter petri dish was filled with 2 mm of water. A Tc-99m saline solution, whose activity was such that the total count rate was below 30,000 counts per second, was added to the petri dish. A 20% energy window centered at 140 keV was used. The petri dish was placed 10 mm away from the collimator surface at the center of the FOV and over 4 million counts were obtained. There was no scatter material present. For each data set, all the counts in the image were summed and the decay-corrected count rate was found. The decay-corrected count rate was then divided by the total activity in the petri dish¹³.

3. RESULTS

The results obtained for the energy resolution, intrinsic uniformity, intrinsic spatial resolution, and system sensitivity tests are summarized in Table 1. The results for the geometric linearity test are displayed in Figure 7. There was a maximum deviation from the theoretical straight line of 0.52 mm. The results of the extrinsic spatial resolution test are

displayed in Figure 8. Both the FWHM and FWTM are displayed, along with the scatter resolution at 10 cm source-to-collimator distance.

4. CONCLUSION

The new LP gamma camera was designed to be a superior alternative to the older thicker camera currently used on the DMT system in terms of overall size and imaging performance. Table 2 summarizes the comparison between the thicker gamma camera's and the LP gamma camera's dimensions and collimator parameters, while Table 3 summarizes the comparison between the cameras' primary imaging performance metrics: energy resolution, intrinsic spatial resolution, and system sensitivity.

The extrinsic resolution of the new camera was also compared to that of the older camera as reported by Gong et al¹⁴. Figure 9 shows the spatial resolution of both cameras plotted versus source-to-collimator distance. The shorter hole length for the LP camera results in slightly worse spatial resolution for large source-to-collimator distance, but provides a factor of ~1.7 improvement in sensitivity.

In addition to improved imaging performance in several aspects, the LP camera provides a thinner profile (8.3 cm thick versus 12.3 cm for the older camera) that permits better camera positioning and greater patient comfort (for example not needing to have the head turned to the side during imaging with the camera above the breast). Also, in addition to the larger nominal imaging FOV of the new camera, the clearer separation of the individual crystal locations in the raw image (see Figure 4) permits crystals all the way to the FOV periphery to be resolved, thus making a larger fraction of the nominal FOV usable. With these improvements in the useful FOV, the new camera permits a reconstructed molecular breast tomosynthesis volume that is similar in size to that of the x-ray tomosynthesis (DBT) component of the DMT scanner. This is important because the smaller of the two volumes sets the upper limit on the amount of breast tissue that can be visualized in the combined structural/functional image.

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Figure 1.

Dual Modality Tomosynthesis system (DMT). X-ray tube, gamma camera, and x-ray detector are on a single gantry arm which rotates around an independent breast compression system. This allows for x-ray and gamma image registration. The gamma camera shown in this photo is the older, thicker one.

Figure 2.

a) H8500C PSPMT with customized circuit board attached. b) Total thickness of final PSPMT module is less than 3 cm.

Figure 3.

a) Camera circuit board showing readout of the 72 X and Y outputs and conversion to ribbon cables. b) Final LP gamma camera assembly. c) LP gamma camera parallel hole collimator.

Figure 4.
Raw flood image used for crystal mapping.

Figure 5.

a) combined and corrected for non-uniformity energy spectrum for Tc-99m source. b) plot of the energy spectrum's peak bin number versus the known photon energies of Am-241 (59.4 keV), Co-57 (122.1 keV), Tc-99m (140 keV), and the fitted linear regression used to calculate the energy calibration factor.

Figure 6.

a) Raw image of the slit before crystal mapping. Here the non-linearities due to the clustering of the detected events can be seen. b) Corrected image after crystal mapping.

Figure 7.
Deviation from linear fit of the image of the slit.

Figure 8.

Results of extrinsic resolution. Source-to-collimator data was taken over 1 to 10 cm from the face of the collimator. Both the FWHM and FWTM can be seen here along with their corresponding resolutions with a 10 cm scatter phantom.

Figure 9.
Comparison of extrinsic resolution between thicker camera and the LP camera.

Table 1

Results obtained for energy resolution, intrinsic uniformity, intrinsic spatial resolution, and system sensitivity.

Parameter	
Energy Resolution:	10.8%
Integral Non -Uniformity:	
UFOV	$6.7 \pm 0.4\%$
CFOV	$6.7 \pm 0.4\%$
Differential Non -Uniformity:	
UFOV -X direction	$4.7 \pm 0.1\%$
UFOV -Y direction	$4.1 \pm 0.6\%$
CFOV -X direction	$4.7 \pm 0.1\%$
CFOV -Y direction	$4.1 \pm 0.6\%$
Intrinsic Spatial Resolution:	
FWHM	2.3 ± 0.1 mm
FWTM	4.2 ± 0.1 mm
System Sensitivity:	240 ± 8 cps/MBq

Table 2

Comparison of thicker gamma camera and LP gamma camera dimensions and collimator parameters.

	Field of View (cm × cm)	Collimator hole length (mm)	Collimator hole diameter (mm)	Collimator septal thickness (mm)
Thicker Camera	15 × 20	21.5	1.9	0.3
LP Camera	20 × 25	18.4	1.9	0.26

Table 3

Comparison of imaging characteristics of the current thicker gamma camera and the new LP gamma camera.

Parameter	Thick Camera	LP Camera
Energy Resolution:	13.5%	10.8%
Intrinsic Resolution: FWHM	2.4 mm	2.3 mm
System Sensitivity:	142 cps /MBq	240 cps /MBq