Abstract—This investigation presents a large area Solid-State Detector (SSD) based on tiled cadmium zinc telluride (CZT) modules for low energy SPECT nuclear medicine applications. The CZT modules used to construct this detector require registered collimation due to the winner-take-all (WTA) readout architecture and charge sharing between pixels. To address this limitation two registered square-hole collimators were fabricated to match the pixel pitch of the SSD. Spatial resolution for each registered collimator was compared to a standard VXGP collimator on a NaI(Tl) EPIC detector.

Studies of 3D Hoffman brain and Jaszczak phantoms show improved image quality. Furthermore, Rollo 2D phantoms show improved contrast and resolution with fewer counts in the SSD images. However, SSD only provides equivalent SPECT imaging performance for cardiac studies.

Comparisons of measured data to simulation models have been used during this investigation to assess SSD detectors and collimation techniques for future camera designs.

I. INTRODUCTION

In recent years dedicated CZT imagers have been proposed [1]-[5]. In this investigation we present results of a large area (20 cm X 40 cm) solid-state detector (SSD) array based on tiled CZT modules.

The imaging performance of the detector has been compared directly to NaI(Tl) scintillation detectors. Both NaI(Tl) and CZT detectors were mounted on a dual detector gantry using the same acquisition system and protocols for all measurements.

The method used in this investigation can be broken down into 4 parts. First, we simulate the imaging performance based on accurate modeling of the module and collimator. Second, we specified and tested the modules that best fit the cost performance model for commercial SPECT cameras. Third, we constructed an imager prototype to verify the models. And forth, this investigation is used to assess potential system designs for SPECT.

A prototype CZT detector using conventional collimation is presented in this paper provides insight to model more complex systems. Pixelated detectors open up a number of possibilities for such system concepts [6][7].

II. MATERIALS AND METHOD

A. Detector Simulations

Analytic models [8] and Monte Carlo simulations [9] were developed to estimate the performance of the detector. The predictions were used to develop algorithms to improve imaging performance at the start of the design. Two of the algorithms developed were super sampling and dead pixel recovery.

1) Super Sampling

Super sampling is a technique designed to reduce planar imaging block artifacts when objects are in close proximity to the detector and to meet the NEMA planar spatial resolution requirements.

The continuous spatial response of the NaI(Tl) scintillator and PMT readout is digitized into a pixel array that must meet at a minimum 1/10 the intrinsic spatial resolution FWHM[10].

Typical NaI(Tl) based cameras have intrinsic resolution approximately 3 mm FWHM with detector sizes of 596.9 mm with 4096 pixels in each dimension. The pixel size of the detector is therefore 0.1457 mm, sufficient for meeting NEMA requirements.

The pixellated CZT detector requires a means to blur the boundaries between pixels without sacrificing resolution. The super sampling technique provides sufficient means of dithering or blurring the pixel-to-pixel response such that the impact on spatial resolution is minimized.

2) Dead Pixel Recovery

There are chances that some pixels in a CZT module may not be usable at all for image detection. Simulations were performed to specify acceptable pixel loss. The tests found that 3% loss began to degrade imaging performance after recovery techniques were employed.

B. CZT Modules

The modules produced and developed by Imarad, an Israeli based CZT supplier, were tested according to Philips specifications. The specifications are based on theoretical performance estimates for energy spectral quality, detection sensitivity, and the results of the simulations.
1) Module Specification

All the Imarad modules tested for this detector were based on the IDEAS ASIC XaIm3.2 [11].

a) Energy resolution

In the case of CZT detectors, the determination of Energy Resolution is complicated by the asymmetry of the peak, due to the difficulty in determining the low-side "toe" of the peak. Several charge transport issues in CZT detectors can cause spectrum tailing. The primary mechanisms for spectrum tailing are hole trapping [12] and charge sharing between the pixels [13]. The spectrum below shows a typical response from a single pixel that is affected by tailing which degrades the energy resolution and photpeak efficiency.

The intrinsic energy resolution should be expressed as a percentage of the peak value and reported clearly whether FWHM, Hide-Side Width at Half Maximum (HSWHM) [14], or Gaussian FWHM (GFWHM) [15] is used.

We used GFWHM to report the energy resolution and to calibrate the detectors. The calibrations provided the KeV/channel correction factor to accurately report the energy resolution.

b) Detection efficiency

For CZT, it is essential to distinguish between the absolute geometric efficiency and the efficiency of the windowed energy peak.

The efficiency of the windowed energy peak describes the total counts that have been measured in the primary peak, where full charge collection has occurred by the given pixel. The other counts in the spectrum represent counts whose charge were not fully collected by the pixel, possibly due to hole tailing, charge sharing, scatter, or recombination.

A measure of the quality of the spectral shape can be found by determining the ratio of the total counts in the peak versus the total counts recorded in the spectrum. While it says little about the absolute efficiency, it provides a measure that may provide insight into the affect of low side spectral tailing. The following definitions help to assess module quality.

Windowed Peak to Spectrum Ratio, WPS, is the ratio of counts in peak area of ± 6.5%Ep energy window (around the peak energy, Ep), relative to total number of counts in the spectrum (integrated from 40 KeV to Ep + 6.5% Ep).

Counting efficiency – Peak, CEP, is the ratio of counts in the peak (Ep +/- 6.5% Ep) relative to total number of counts crossing the plane of the crystal face during the counting time.

Counting efficiency – Spectrum, CES, is the ratio of counts in the spectrum relative to total number of counts crossing the plane of the crystal face during the counting time. The CES performance metric is calculated from CEP and WPS measurements.

c) Dead Pixel

Dead pixel is defined as noisy or unresponsive pixel with a Windowed Peak to Spectrum Ratio (P to S) below 35%, CES-Peak below 25% or Energy Resolution (GFWHM) above 13%.

The number of dead pixels should be no more than 3 per module and no more than two can be adjacent, which is less than the maximum of 3% dead pixels found in simulations to ensure acceptable image quality.

III. TEST RESULTS

A. Module Test Results

Figure 2 shows distributions for WPS, and CEP, and CES using Co-57 give a quick assessment of the modules independent of the imager. Slight variations should be considered when viewing the distributions. For instance 3 modules have less counts in the CEP distribution. This is due to measurement error by incorrectly placing the source in the station.

By controlling the temperature, shaping time, and bias voltages it was demonstrated that module performance for energy resolution could improve up to 20% with a 5% increase in collection efficiency.

1) Temperature and shaping time

Module performance is degraded by thermal noise in the front-end circuit and by increasing the leakage, or dark, current [16]. These factors limit the peaking times to a few microseconds [17,18]. Surface and bulk leakage current can be reduced significantly by dropping the temperature 10°C [19].

The maximum shaping-time setting was 812 ns with the default set to 487 ns, best results (Module GFWHM average=3.75%, min=2.90% and max=5.24%) were found at 673 ns with HV set to −750 and with temperature at the ASIC and CZT crystal at 6°C and 18°C respectively.

2) Bias voltage

In figure 2, the detector HV bias was varied from -400V to -750V. Test results show the energy resolution is optimum at -600V. The windowed counts do increase with increased HV
bias suggesting that the sensitivity can be improved by increasing high voltage.

![Graph showing impact of high voltage bias](image)

Figure 2 Impact of High Voltage Bias

**B. Description of SSD imager**

The SSD is composed of 60, 5 mm thick, CZT detector modules with 256 pixels each, the modules produced by Imarad with 2.46 mm pixel pitch. The 60 modules were arranged 5 rows by 12 columns. Each row had a separate readout channel that was processed by a Xilinx XC2V2000 FPGA. The processing including gain-offset adjustment for each pixel, dead pixel recovery, super-sampling, and uniformity corrections. The calibrations were performed in a single step using specifically design flood phantom to generate sufficient statistics for corrections of all 15,360 pixels in a short time period, <1 hour.

**C. System Performance**

The Forte gantry was used to compare an EPIC NaI(Tl) detector with the CZT SSD. The NaI(Tl) detector used a VXGP for all comparisons with the SSD detector. The SSD detector used two types of collimators. Each registered collimator consists of bores of lengths 35 mm and 45 mm having a hole size of 1.71 mm and 0.75 mm Pb-septa thickness.

1) **Energy Resolution**

The SSD consisting of 15,360 pixels had a measured energy resolution of 6.7% FWHM versus 9.0% for the 3/8" or 9.5 mm thick NaI detector. The energy resolution for the entire SSD array with registered collimator was measured using a flood phantom with Tc-99m; the peak was calibrated using the lead x-ray escape peak.

In figure 4, the lower peak is shows the Pb x-ray from the lead in the collimator and it is used to calibrate out offsets for the energy resolution calculation.

The impact of collimator alignment was also studied as it relates to spectral quality. The collimators were varied in 1 dimension and 2 dimensions for both 35 mm and 45 mm collimators. The spectral tailing is observed when the collimators are misaligned with the 35 mm collimator tailing significantly more than the 45 mm. This is most likely due to the wider acceptance angle of the shorter bore.

![Composite energy spectra](image)

Figure 4 Composite energy spectra of CZT imager.

2) **Sensitivity, Spatial Resolution and Contrast**

In table 1, case 2 used the same exact collimator and source placed onto a NaI based detector and acquired under the similar conditions, 13.5% window for CZT and 20% window for NaI to compare with case 1 which we compared to theoretical efficiency of the CZT detector.

<table>
<thead>
<tr>
<th>Comparison Cases with NaI Detector using Tc-99m</th>
<th>Intrinsic CZT</th>
<th>CZT/35</th>
<th>CZT/45</th>
</tr>
</thead>
<tbody>
<tr>
<td>1) CZT (13%) counts to NaI (20%) counts</td>
<td>56.8%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2) Same collimator used on both</td>
<td>67%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3) Measured against NaI/VXGP</td>
<td>74%</td>
<td>44%</td>
<td></td>
</tr>
</tbody>
</table>

Table 1 Relative Sensitivity of CZT to NaI(Tl).

Theoretical detector efficiency of 5 mm thick CZT is 76% efficient for the effective absorption of 140 KeV photopeak gamma photons. Assumptions are that 50% of Compton
interactions recombine with 5% scattering out of the pixel. NaI(Tl) detectors are 86% efficient, making the relative efficiency to be 75.5% divided by 86% equaling 88%. Registered collimation improves the relative sensitivity by 10% from a 30% discrepancy from the theoretical for the intrinsic case. We suspect that the 20% difference is due to hole tailing and incomplete charge collection in the photopeak window.

Since collimator designs are different from VXGP to 35mm and 45mm we can tradeoff resolution for sensitivity. Cases 3 shows that the sensitivity of the CZT/35REG detector can be improved while table 2 shows that the resolution of the CZT/35REG is still better than the NaI/VXGP detector. The collimators were constructed of lead with manufacturing cost as a key requirement. It is feasible to build more efficient collimators of similar construction that are optimized for higher sensitivity. Registered collimation does improve the relative sensitivity of the detector. This indicates that charge sharing impacts intrinsic detector performance by 10% for 140 KeV photons as shown in table 1.

Line source resolution measurements were performed for the CZT/35mm, CZT/45mm and NaI/VXGP combinations. Four line sources were placed at 0, 5, 10 and 15 cm from the detector surface and parallel to each other. The results are shown in table 2.

<table>
<thead>
<tr>
<th>Source Distance (cm)</th>
<th>CZT/35 FWHM (mm)</th>
<th>CZT/45 FWHM (mm)</th>
<th>NaI/VXGP* FWHM (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>2.8</td>
<td>2.8</td>
<td>4.8</td>
</tr>
<tr>
<td>5</td>
<td>4.3</td>
<td>3.8</td>
<td>6.2</td>
</tr>
<tr>
<td>10</td>
<td>6.8</td>
<td>5.6</td>
<td>7.9</td>
</tr>
<tr>
<td>15</td>
<td>9.4</td>
<td>7.5</td>
<td>9.8</td>
</tr>
</tbody>
</table>

Table 2 Spatial resolution as function of distance

*VXGP cover adds 8.5 mm to source distance

To minimize the effect of collimator hole shape, the line source images were measured several times with the different orientation angle. The orientation angles used were 7.5, 15, 22.5, 30, 37.5 and 45 degrees relative to the camera rotation axis. The final FWHM at each source distance is an average of the FWHMs at different orientation angles.

D. Phantom Studies

A planar phantom study is presented here with three sets of SPECT phantom studies along with the imaging and reconstruction protocols. The Jaszczak and brain studies are high-count image sets to assess the imager performance with reduced statistical noise. The CZT used the 45 mm Registered collimator for both studies. In both studies, the amount of counts in the NaI(Tl) was approximately double that of the SSD. The improvement is even more significant as counts increase showing the inherent advantages of SSD.

1) Rollo Phantom

Static planar acquisitions of Rollo contrast phantoms were taken at 1-minute intervals to study the statistical nature of the new imagers. Although the sensitivity of the SSD was less than NaI, the resolution and contrast were noticeably improved, see figure 5.

![Figure 5 Rollo Contrast Images](image_url)

2) Jaszczak Phantom

Jaszczak phantom images were acquired simultaneously for both the detectors with 128 stops in 360 degrees and 60 seconds per stop. Post-acquisition uniformity correction and decay correction was applied for both the CZT and the NaI acquisitions.

A zoom factor of 1.06 was applied to the CZT reconstruction so that the resulting pixel size was the same as the NaI detector. Attenuation correction was applied using the post-reconstruction Chang’s method.

<table>
<thead>
<tr>
<th>Number of Projections</th>
<th>128</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time per projection</td>
<td>60 sec</td>
</tr>
<tr>
<td>Starting Activity</td>
<td>70 mCi</td>
</tr>
<tr>
<td>Reconstruction</td>
<td>CZT Imager, NaI-VXGP</td>
</tr>
<tr>
<td>Pixel Size (nm)</td>
<td>2.46, 2.33</td>
</tr>
<tr>
<td>Reconstruction</td>
<td>FBP, FBP</td>
</tr>
<tr>
<td>Filter Type</td>
<td>Gaussian, Gaussian</td>
</tr>
<tr>
<td>Cutoff/Order</td>
<td>0.2/1.5, 0.2/1.5</td>
</tr>
<tr>
<td>Zoom (ASP)</td>
<td>1.06, 1.00</td>
</tr>
</tbody>
</table>

Table 3 Acquisition and Reconstruction parameters for Jaszczak phantom
The improvement is highly appreciated for CZT, as shown in figure 6 both pies and spheres are clearly observed. The smallest sphere can be observed; the 5th ball is missing from the phantom.

3) Hoffman 3D Brain Phantom
The brain phantom benefits the most from the CZT detector. The better resolution and contrast are evident in figure 7. Care was taken to match the plot profiles as closely as possible. The CZT profile in orange was adjusted to match the NaI profile in white by scaling the difference in average counts.

<table>
<thead>
<tr>
<th>Number of Projections</th>
<th>128</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time per Projection</td>
<td>60 sec</td>
</tr>
<tr>
<td>Starting Activity</td>
<td>30 mCi</td>
</tr>
<tr>
<td>Reconstruction</td>
<td>CZT imager NaI-VXGP</td>
</tr>
<tr>
<td>Pixel size (mm)</td>
<td>2.46 2.33</td>
</tr>
<tr>
<td>Reconstruction Method</td>
<td>FBP FBP</td>
</tr>
<tr>
<td>Filter Type</td>
<td>Gaussian Gaussian</td>
</tr>
<tr>
<td>Cutoff/Order</td>
<td>0.2/1.5 0.2/1.5</td>
</tr>
<tr>
<td>Zoom</td>
<td>1.06 1.00</td>
</tr>
</tbody>
</table>

Table 4 Acquisition and Reconstruction parameters for brain phantom

The improvement in cardiac image is less significant as shown in figure 8 for Tc-99m when compared to the Jaszczak or brain phantoms. The improvement shown in the thickness of the wall with less intensity in the basal areas in the CZT image indicates the rejection of scatter, in turn, shows more attenuation.

4) Cardiac Phantom
An anthropomorphic thorax phantom with cardiac insert was used for cardiac imaging performance evaluation. The experiments included Tc-99m studies with normal and abnormal hearts, Tl-201 studies and investigation of down scatter contamination from Tc-99m (140 keV) photons to the lower Tl-201 (70 keV) window.

One application of simultaneous dual isotope (SDI) imaging is for cardiology, which uses two isotopes Tc-99m (140 KeV) and Tl-201 (70 KeV) to assess perfusion of the myocardium.

We measured the amount of signal down scattered into the Tl-201 acquisition window using a water phantom filled with Tc-99m only. Then adding Tl-201 to measure the signal with contamination from Tc-99m. The premise of better energy resolution does not evident in this case. However, the fraction of contamination of the 140 KeV scatter into the 70 KeV window improves as the acquisition window width for CZT narrows. We conclude that spectral tailing limits the performance of the CZT detector for such SDI applications.

Table 5 Acquisition and Reconstruction parameters for Cardiac phantom
IV. DISCUSSION

We present a large area CZT detector and methodology to compare two systems in a clinically relevant way.

In general, the images show that the CZT detector performed as well or better than the NaI detector even with lower sensitivity. However, the NaI detector marginally outperformed CZT detector in the case of SDI due to spectral tailing of CZT detector. Improved sensitivity and SDI are key requirements to be considered with system cost to develop a successful product.

The spectral tailing can be attributed to charge transport issues of a pixel array in a slab of CZT, these have been investigated [20][21]. The energy resolution and count efficiency were simulated as a function of semiconductor thickness and pixel pad size. Narita confirmed these simulations with respect to thickness and pad size [22]. Further confirmation is shown in table 1 highlighting the advantages of registered collimation and consistency with the individual module test results shown in figure 1. Further investigations are needed to understand the spectral tailing and sensitivity issues observed.

King [23] identifies 3 sources of image degradation in SPECT. The first degradation of SPECT is attenuation. CT with SPECT attenuation correction techniques will reduce this impact considerably. The second degradation, scatter, is a result of the finite energy resolution of the imaging system. The third degradation problem of spatial resolution is due to depth-dependent blur and position estimation. Our goal is to use the tools developed during this investigation to overcome the third problem while engaging our suppliers to solve the second.

The construction of a conventional large area camera with a pixellated CZT detector and Anger NaI(Tl) detector allow us to compare and separate the SPECT components to a basic level. This gives us confidence that our simulations are accurate. Simulations and image quality assessment tools developed show that sensitivity can improve significantly without loss of spatial resolution performance.

By using both high statistic and lower statistic images, see figure 5, we assess the robustness of the imaging system and begin to understand the noise properties of the system before reconstruction [24]. As we improve reconstruction techniques with these tools, we have additional opportunities to improve upon the classical sensitivity-resolution tradeoff in SPECT.

We made a good start at understanding the advantages of pixellated solid-state detectors as compared to NaI(Tl) detectors. The tools established with this investigation will be used for continuous search and evaluation of new ways to improve SPECT image quality.

V. ACKNOWLEDGMENT

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VI. REFERENCES

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