Published in final edited form as:

IEEE Trans Nucl Sci. 2012 August 30; 59(5): 1831–1840. doi:10.1109/TNS.2012.2210908.

Development of a Cost-Effective Modular Pixelated NaI(Tl) Detector for Clinical SPECT Applications

Mike Rozler [Member, IEEE], Haoning Liang [Member, IEEE], and Wei Chang [Member, IEEE]

Department of Diagnostic Radiology, Rush University Medical Center, Chicago, IL 60612 USA

Mike Rozler: michael_rozler@rush.edu

Abstract

A new pixelated detector for high-resolution clinical SPECT applications was designed and tested. The modular detector is based on a scintillator block comprised of $2.75 \times 2.75 \times 10 \text{ mm}^3$ NaI(Tl) pixels and decoded by an array of 51 mm diameter single-anode PMTs. Several configurations, utilizing two types of PMTs, were evaluated using a collimated beam source to measure positioning accuracy directly. Good pixel separation was observed, with correct pixel identification ranging from 60 to 72% averaged over the entire area of the modules, depending on the PMT type and configuration. This translates to a significant improvement in positioning accuracy compared to continuous slab detectors of the same thickness, along with effective reduction of “dead” space at the edges. The observed 10% average energy resolution compares well to continuous slab detectors. The combined performance demonstrates the suitability of pixelated detectors decoded with a relatively small number of medium-sized PMTs as a cost-effective approach for high resolution clinical SPECT applications, in particular those involving curved detector geometries.

Index Terms

Biomedical imaging; gamma camera; single photon emission computed tomography

I. Introduction

For many decades the Anger camera [1] has been the workhorse of single photon emission computed tomography (SPECT). Anger's elegant design owes its longevity in a big part to the cost-effective and robust way in which it meets clinical SPECT's high stopping power and moderate resolution requirements. Slab detectors consisting of continuous NaI(Tl) slabs decoded with arrays of large conventional (single-anode) photomultiplier tubes (PMTs) are inexpensive to manufacture and have relatively few output channels, thus placing only moderate demands on the processing electronics. Today's state-of-the-art clinical SPECT dual-head gamma cameras typically use large, 76 mm diameter, PMTs and yield detector intrinsic spatial resolution (ISR) of 3.8 mm [2] averaged over the detector's useful field of view (UFOV). Improvement/of detector ISR is problematic for a continuous scintillator because the thick slabs required to achieve high stopping power lead to excessive scintillation light spread. Another limitation of the Anger camera is the sizable “dead” space near the edges, which is unusable over a distance of several centimeters. This can be dealt with cost-effectively in conventional SPECT systems by enlarging the size of the crystal slabs of the cameras beyond the required UFOV.

In addition to its contribution to system resolution performance, detector ISR is important in cameras utilizing projection minification to increase the number of simultaneous projections acquired by the system. This is the case for C-SPECT [3], the dedicated cardiac platform

being developed by our group. The system, the design concepts of which are illustrated in Fig. 1, employs inverse fan-beam collimation which minifies the projections by about a factor of two transaxially. This permits the simultaneous acquisition of a large number (13–14) of non-overlapping projections, thus increasing the system geometric efficiency (SGE), as well as reducing (or eliminating) the number of patient rotations required for adequate sampling. Finally, novel cardiac SPECT cameras [4], [5], as well as C-SPECT, boost SGE by utilizing curved detector geometries to minimize distance and reduce or eliminate gaps in detector coverage. Since curved single-piece scintillator crystals large enough to accommodate a patient's torso and maximize sensitivity for cardiac imaging have never been developed, modular assemblies are necessary. Thus, any “dead” space at the edges of the modules, if present, would be a serious problem that would render a significant fraction of the total detector area ineffective for imaging.

Our approach to cost-effectively addressing these issues in a large curved detector is the same one used in positron emission tomography (PET): modular pixelated scintillation detectors [6], [7]. Pixelated detectors employ scintillator crystals divided into small pixels separated by thin reflectors. The resulting light-piping effect controls the light spread to achieve high ISR. “Dead” space, unacceptable in modular curved detectors, can also be effectively reduced or eliminated.

The high ISR of pixelated scintillators can be used to achieve high detector resolution by combining small pixels with multi-anode PMT's (position sensitive PMTs or PSPMTs), which yield finer spatial sampling of scintillation light [8]. While this approach has shown excellent performance for applications requiring a relatively small detector area, such as pre-clinical imaging or scintimammography [9], [10], the high cost of PSPMTs poses a problem when a large-area detector system is required. To adapt pixelated scintillator technology to meet general clinical SPECT's stringent cost constraints, we propose the use of relatively large single-anode PMTs, thus containing the individual cost and number of PMTs, signal processing channels, and associated hardware. This detector approach may also find applications in development of other systems requiring large curved detectors, such as brain-SPECT cameras. In this paper we evaluate the design options and performance potential of this approach using a small test detector. We have also develop a method to objectively evaluate detector performance in terms of event positioning errors on a pixel by pixel basis and to present the results in both a quantitative and qualitatively intuitive form.

II. Materials

The predecessor to our test detector is the NaI(Tl) PET detector developed by the University of Pennsylvania group [11] in collaboration with Saint-Gobain (Saint-Gobain Crystals, Hiram, Ohio). That detector used $4 \times 4 \times 30$ mm³ pixels and is read out by a hexagonal array of 39 mm diameter PMTs. Our goal is to investigate this approach's performance potential for SPECT applications and push it further by decreasing the pixel size, thus improving ISR and linear sampling. Another aim is to increase the PMT size, thus reducing the number of PMTs and, therefore, system cost. We also evaluate several options for improving detector performance: exit window edge treatment and PMT type variations.

As shown in Fig. 1(c), the basic unit of C-SPECT's detector is a detector module decoded by 3 columns of a hexagonal array of 51 mm single-anode PMTs. Each PMT of the two edge columns is shared with the adjacent module. Each module's approximately 90 mm frontal transaxial width is constrained by the PMT diameter and the hexagonal PMT pattern. Axially, the modules are 200 mm long, requiring a total of 121 PMTs for the entire 13-module detector system.

We worked with Saint-Gobain, which built a small test section of C-SPECT's planned detector system. The detector is based on $2.75 \times 2.75 \times 10 \text{ mm}^3$ NaI(Tl) pixels. To emulate a section of the detector arc, two slightly tapered modules are angled at 15° with respect to each other, matching the average curvature of C-SPECT's detector. Each module contains a $90 \times 51 \text{ mm}$ NaI(Tl) block comprised of a 30×17 array of pixels separated by a 0.25 mm-thick white diffuse reflector, resulting in a 3.0 mm detector pitch. The detector photo and a cutaway view are shown in Fig. 2. The 90 mm transaxial length of the modules matches C-SPECT's future detector modules. Axially, the test module is significantly shorter; the modules can be easily extended in that direction, along with the PMT array.

Since NaI(Tl) is hygroscopic, hermetic sealing is a requirement. Pixel blocks have to be sealed with glass exit windows in an aluminum housing. To maintain the seal, the edges of the housing rise above the pixels; by blocking scintillation light, non-transparent edges can degrade the performance of the detector for pixels close to the edge. One of our goals is to evaluate the use of black vs. white epoxy at the window-aluminum seals. Intuitively, use of a white epoxy can allow more scintillation light to reach the PMTs, maintaining energy resolution. Use of black epoxy, on the other hand, would prevent most unwanted reflection, at the cost of losing some scintillation photons and energy resolution.

The two modules of our test detector are sealed with 6 mm-thick glass exit windows in a single common aluminum housing, allowing us to evaluate a third edge treatment option: a clear epoxy is used to seal the joint between the glass windows of the two modules in order to allow light propagation across the interface. The epoxy used at the joint edges has to be chosen more for its mechanical strength than optical properties. As a result, light transmission across this interface appeared marginal upon visual inspection. The common housing also permits a slight reduction of the critical inter-module gap to 0.9 mm from the 3.0 mm gap mm required for separate housings. At the other edges, the exit windows are sealed to the housing using black epoxy for one of the modules (referred to as “black” module), and white epoxy for the other (referred to as “white” module). The hermetic seal and the housing attain their mechanical strength from the 6.4-mm wide Al shoulder and the 7.4 mm-thick Al wall on the two axial edges. A shoulder can only be used at these axial edges, since the detector arc is formed by joints at the transaxial edges.

A hexagonal array of 12 PMTs was used for signal readout. A 3-piece modular UVT acrylic (Saint-Gobain BC-800 material) lightguide, shown in Fig. 3, was used to spread the scintillation light from the detector to the PMTs. The lightguide thickness, which has to vary slightly due to the detector curvature, was chosen based on decoding maps obtained with flat lightguides.

In all experiments, Dow Corning Q2-3067 optical couplant was used for coupling of the detector modules, lightguide and PMTs. The set-up is shown in Fig. 3. Two types of Hamamatsu 51 mm diameter single-anode PMTs were used in our experiments: a standard PMT, R1306-19, with a nominal quantum efficiency (QE) of 25%, and the more expensive R6231-100 with a higher QE of 33%. These are referred to as SQE and HQE, respectively. Intuitively, the HQE PMTs would be expected to yield some performance advantage due to their superior light response efficiency. Because of the HQE PMTs higher cost (around \$450 per HQE PMT vs. \$300 per SQE PMT), we are particularly interested in evaluating their selective use at all the joints between the modules in a system, due to the scintillation detector's poorer performance at the edges, as described above. In the configuration used to compare the two detector modules, the two PMTs located above the joint between the modules were HQE PMTs; the other ten were SQE PMTs.

III. Method

A. HQE PMT Evaluation

To evaluate the impact of using the HQE PMTs, we tested four PMT combinations. These measurements were conducted by scanning the black module only, since that module showed better positioning performance. These configurations are shown in Fig. 4. Only the 7 PMTs coupled to the black module and at the module joint were varied. The 5 PMTs coupled to the white module remained SQE.

B. Data Acquisition

The DAQ system used to acquire PMT signals is schematically shown in Fig. 5. In all experiments, list-mode digital data were collected, with each event packet containing the peak amplitude for all PMTs. Amplifier shaping time was set at $1.5 \mu\text{s}$. A wide hardware energy window was used, with appropriate narrower digital energy window applied in post-processing.

C. Beam Scan

The most important measure of detector performance is the difference between actual and detected event positions. We evaluated positioning performance of the test detector directly by performing a pixel-by-pixel beam scan. Since the actual location of the pixel being illuminated is known, decoding errors can be measured directly for each event. The beam scan approach was used to evaluate positioning accuracy directly by Dahlbom and Hoffman [12]. In their work, the fraction of correctly decoded events for selected pixels of a small PET detector is presented. We have extended this approach by calculating average positioning errors in both dimensions for all pixels. The event positioning errors can then be analyzed pixel by pixel, regionally, or globally, and presented in numerical and/or graphical forms.

We used an XY translation stage to scan a collimated Tc-99m beam source across the detector, aiming at one pixel at a time, as shown schematically in Fig. 6. Measurements of the detector were performed with the beam centerline at 7.5° from normal incidence when evaluating both modules at the same time, or at normal incidence when evaluating one module. Care was taken to align the XY stage with the detector to ensure the beam is aimed the same way at each pixel. The beam centerline was always aimed at the pixel center. The detector was enclosed in a lead-shielded box to reduce the background. A typical starting count rate in our experiments was around 2,000 events/sec, with the background rate of about 50 events/sec. In all experiments, 5,000 events were obtained for each scan position. With most scans taking several hours, the count rate would typically drop by a factor of around two from the beginning to the end of the scan.

D. Decoding Scheme

Once the complete data set (5000 events per pixel) is acquired, the decoding is performed offline in Matlab. Each event acquired by our DAQ system contains signals from each of the 12 PMTs. To improve high-count-rate performance in the future C-SPECT system, local module triggering will be implemented in the DAQ system. For any single event, a fast analog signal will be used to quickly compare the signals from PMTs of neighboring modules to allow acquisition of signals from PMTs coupled to only one, the correct, module [13]. To emulate this local triggering approach in our decoding scheme, each event's signal sums of the three central PMTs of the black and the white modules were compared. Only the signals from the 7 PMTs of the module with the larger sum of the 3 PMTs in the middle column were used for decoding.

Following Dahlbom and Hoffman [12], we also generate lookup tables (LUT), which relate coordinates calculated by Anger logic from measured PMT signals with the appropriate detector pixel. Since a beam scan was being performed for the purpose of detector evaluation anyway, we used the same beam scan to generate the LUTs. The first 1000 of the 5000 events obtained for each pixel are calibration events used to generate the LUTs, while the other 4000 are considered data events, used for evaluating the positioning accuracy.

The calibration procedure in which the LUTs are generated is performed as follows. The positions of calibration events are first calculated by Anger logic. The position space spanned by all positions calculated for calibration data is divided into 200×200 position bins. A $200 \times 200 \times 3$ position look-up table (PLUT) is then generated. Each position bin of the PLUT contains the three likeliest pixels of origin for this bin, in the order of their likelihood. To do that, all calibration events for which the calculated position lies in the bin are sorted by their pixel of origin (the pixel the beam was aimed at when the event was acquired). The three pixels of origin that most often result in event positions lying in the bin are recorded in the PLUT. Each detector pixel, on the average, is associated with 8 position bins in the PLUT. Additionally, a local energy look-up table (ELUT, one entry per pixel) is generated during calibration. This ELUT contains the peak PMT signal sum for events in each pixel (the terms “energy”, “pulse height”, and “PMT signal sum” are used loosely and interchangeably throughout this paper). The PMT signal sum for every pixel is determined by fitting the photopeak energy spectrum of the 1000 calibration events to a Gaussian.

For the pixel assignment procedure, the position of each event is calculated by Anger logic. The event is then assigned to the most likely pixel of origin based on the PLUT; the event is rejected if the calculated position lies outside of the range of the PLUT. A local energy window from the ELUT is then applied to test whether the event's PMT signal sum lies within $\pm 10\%$ of the PMT sum associated with the likeliest pixel. If the check is successful, the event is assigned to this likeliest pixel. If the event fails the energy check, the check is performed for the second likeliest pixel, and the third likeliest pixel if it fails again. If the energy test is failed for all three likeliest pixels, the event is discarded.

Many of the parameters used in the decoding scheme, such as number of PLUT bins, energy window width, and number of energy window test attempts were chosen by trial and error to balance improvement in positioning accuracy and reduction of event rejection.

The positioning error for any single evaluation event is expressed as an integer number of pixels, along each of the two dimensions: transaxial (X-error) and axial (Y-error). The quadrature sum of the two is referred to as “R-error”. Mean errors $\overline{R_{er}}$, $\overline{X_{er}}$ and $\overline{Y_{er}}$ are calculated for every pixel by averaging errors over the accepted events acquired when the beam was aimed at the pixel. For the X- and Y-errors, absolute values are taken before averaging.

All the accepted events are also used to generate a traditional decoding map: a 2D histogram of event positions as calculated by Anger logic. We used 256×256 bins for all of the decoding maps, which are presented throughout this work to help qualitatively visualize resolvability of pixels. Finally, pulse height resolution, σ_E/E , is computed for each pixel by performing a Gaussian fit to the energy spectrum from the 5,000 events acquired with the beam aimed at the pixel. The width σ_E is measured as the full width at half maximum (FWHM), and the PMT signal sum E is the peak value of the fit.

For the simultaneous measurement of the two modules, a decoding map acquired with a flood field was also obtained, using a distant source. The total number of events acquired to generate the flood field map was equal to the number of events detected for the beam scan

map ($4000 \times 60 \times 17 = 4,080,000$ events). All detected events are included in the flood decoding map.

IV. Results

A. General Performance

The decoding map of the test detector is shown in Fig. 7. Qualitatively, the pixels are well resolved, with poorer pixel separation at the edges and near PMT centers. The black module shows better pixel separation and smaller pincushion distortion, compared to the white module.

The histogram of R-errors, averaged over all pixels of both modules and all events is shown in Fig. 8. Since the errors are expressed as the number of pixels, the errors are “quantized”. For any given target pixel, misdecoding to one of the four nearest neighbors produces an R-error of 1 pixel, to one of the four “diagonal” neighbors—an R-error of 1.41 pixels, etc. In the histogram presented, for every event assigned to the correct pixel, an average of 0.42 events are misdecoded by one pixel, an average of 0.1 events are misdecoded by 1.41 pixels, and so on. Since majority of the pixels in the module have 4 neighboring pixels, for every correctly decoded event an average of 0.105 events are misdecoded to each of the four nearest neighbors and an average of 0.025 events are misdecoded to each of the four “diagonal” neighbors. As would be intuitively expected, the error distribution falls rapidly with distance, flattening out to a low value by the distance of about 6 pixels from the actual event. Although the distant events are relatively few in number, they continue to make a contribution to the average error since the contribution is weighted by the distance. We attribute these large errors to the background events, while the errors at closer distances are actually related to the detector ISR. This is illustrated in Fig. 9, which shows the dependence of the R-error (averaged over both modules) on event count rate (measured at the beginning of the scan) for eight sequential full module scans. One of the three plots in the graph shows the error averaged over all accepted events. The other two are averaged only over the events with R-error of less than or equal to 3 pixels, and less than or equal to 6 pixels, respectively. Without cutting off the long distance error, the data is strongly dependent on the count rate, since the background has more impact at lower count rates. The data in this work is presented with setting the R-error cutoff at 6 pixels, to be more representative of detector ISR, because cutting the error off at smaller values may undercount “legitimate” ISR-related errors. For reference, we anticipate the actual count rate for each C-SPECT module to be around 2,000–5,000 events/sec.

In order to present the numerical error data in an intuitive form, we show maps of mean errors averaged over all events for each pixel. The map of $\overline{R_{er}}$, magnitude of the decoding error (in number of pixels) is shown in Fig. 10(a). The maps of $\overline{X_{er}}$ and $\overline{Y_{er}}$ the errors in the transverse (X) and axial (Y) directions, are shown in Fig. 10(b) and (c). Fig. 11(a) shows the map of pulse height resolution, E/E . The map of the pulse height E is shown in Fig. 11(b). Although E is not a standard detector metric, it is a useful way to visualize relative local scintillation light collection in the detector. Finally, Fig. 12 shows the map of the fraction of events aimed at each pixel that are accepted. In an operational system, this map would be used to provide uniformity correction for each pixel

The values of mispositioning errors, R_{er} , $\overline{X_{er}}$ and $\overline{Y_{er}}$ averaged over each of the two modules, are presented in Table I, along with the pulse height resolution (E/E) and acceptance fraction (f_a). All metrics are dimensionless, except pulse height resolution, which is in %. Two other metrics are included in this and other tables, without a map shown. One of them is f_c —the fraction of events decoded to the correct pixel. The other, f_{nm} is the

fraction of events decoded to the correct pixel, its nearest neighbor or the diagonal neighbor pixel (3×3 square around the event location). In addition, the detector metrics are also presented broken down over seven regions of each detector module in Table III of the Appendix.

B. Impact of PMT Type

The measurements conducted to evaluate the impact of PMT type were performed on the black module at normal average incidence. The 5 SQE PMTs left unchanged on the white module play no role in decoding, since their signals are rejected by the emulated local triggering, except for background events. Fig. 13 shows the decoding maps obtained for the PMT configurations described in Section III-A. The R-error maps and the pulse height resolution maps are shown in Figs. 14 and 15, respectively. The values of the detector metrics averaged over the modules are summarized in Table II. Table IV in the Appendix contains the metrics broken down by the 7 local regions. While positioning performance and pulse height resolution improve with the increased use of HQE PMTs, we have no explanation the reduction in acceptance fraction in the 7-HQE configuration. All data show that there is no performance advantages that can be gained with the use of common housing, compared to non-transparent black edge treatment.

C. Energy Spectra

Fig. 16 shows energy spectra obtained from 4 representative detector pixels. We show the spectra from the 7 SQE configuration to put the pixels on more equal footing. Each spectrum contains 5000 events. The $\pm 10\%$ energy window is marked for reference. The fraction of events lying within each pixel's energy window is consistent with the fraction of events accepted by our positioning algorithm, when taking account of the fact that some events are mispositioned. We do not have an explanation for the poor energy resolution consistently observed in the top right corner of the detector (location 4), and consider it to be an anomaly.

V. Discussion

We tested a small two-module section of modular pixelated detector based on $2.75 \times 2.75 \times 10\text{mm}^3$ NaI(Tl) pixels decoded by an array of 51 mm diameter single-anode PMTs. To investigate performance and design features of the modules, we performed a comprehensive analysis based on quantitative metrics for pixelated detector performance. The detector shows good event positioning performance, achieving correct pixel identification for 60–72% of all events, depending on the measurement configuration. This Corresponding energy resolution averaged over a single module ranged between 9.1 and 10.5%.

Black edge treatment provides higher positioning performance than white edge treatment, although at the expense of a small pulse height resolution loss. For the 10 SQE/2 HQE PMT configuration, the black module's mean R-error is 0.41 pixels, compared to 0.52 pixels in the white module. The white module's 10.0% average pulse height resolution is only marginally better than 10.2% measured for the black module. The common housing with transparent epoxy at the joint of two glass windows has not led to any improvement in detector performance and will not be used in further design, since it reduces system modularity. Instead, the future detector will be based on separate single flat sealed modules.

HQE PMTs provide a measurable but incremental advantage over the SQE PMTs in both event positioning and energy resolution. Knowledge of the specific improvement yielded by the use of HQE PMTs in every region of the detector enables a meaningful cost-benefit analysis of the use of these costlier PMTs. Traditionally, pixelated detector positioning performance is evaluated by using peak-to-valley ratios to indirectly gauge the positioning

errors from the overlap of scintillation light profiles from adjacent pixels. Although this ratio can be used for comparisons of detector designs, we consider the knowledge of absolute position errors, in both global, regional and local terms, as well as quantitative and qualitative aspects, to be valuable. Use of a beam scan to evaluate the performance of the detector by directly measuring decoding errors yields a wealth of detailed quantitative information. This is important at the development stages of a detector system and allows objective comparisons of design variations. It should also allow realistic modeling of the detector to guide its improvement, as well as to aid in development and optimization of the complete imaging system.

In general, the use of the ELUT in addition to Anger logic is the smallest possible step towards using higher dimensional signal information for event positioning. The algorithm used in this work, based on (2+1) -dimensional LUT, should present no problems for real-time decoding of events. Taking the approach to its full extent would involve using all possible combinations of PMT outputs to create a very high-dimensional LUT. This is the essence of maximum likelihood (ML) positioning, as introduced by Gray and Makovski [14] and implemented by the University of Arizona group for modular slab detector systems with 9 PMTs per module [15]. ML positioning is a deep research subject in its own right, and it is possible that a pixelated detector of the type presented here could benefit from such algorithms.

A practical concern for a detector system is the ease of calibration. In this work, pixel-by-pixel beam scan calibration was convenient since the beam scan was performed anyway to evaluate positioning accuracy. While it can certainly be performed during commissioning of the detector, beam scan calibration is not practical for a routine quality control calibration of a full-scale system. Dahlbom and Hoffman [12] describe a flood scan calibration approach, based on generating the position LUT using regions defined around each pixel in the flood field decoding map. We plan to implement such an approach, but it is likely that it will result in slightly degraded performance compared to a calibration based on knowing each event position accurately provided by the beam scan. A method for a practical and fast calibration could involve a lead calibration plate with multiple holes, exposing only a subset of the detector pixels. For example, only a quarter of the pixels could be exposed at a time, so that no neighboring would be exposed simultaneously. The calibration could then be performed using a flood field in four steps, one quarter of the pixels at a time, avoiding pixel overlaps. In essence, this can be considered to be a parallelizing of a beam scan. Such partial exposure techniques have also been used in the linearity and energy calibration of commercial Anger cameras. A hybrid calibration approach, utilizing a flood field calibration combined with a beam scan for the more poorly resolved edge regions, is also a possibility.

Compared to a continuous slab, the effective stopping power of a pixelated detector is reduced due to the presence of the reflector material between the pixels. Given the present detector's pixel and reflector dimensions, its packing fraction or stopping power at normal incidence is 84% of that of a continuous slab of identical thickness. In fan beam collimation, a large fraction of the photons is incident obliquely; thus the stopping power of the detector when used with C-SPECT's slit-slat collimator will be higher. The stopping power loss in the axial direction will also be smaller due to the presence of the collimator's slats, which will be aligned with the detector's reflectors.

It may be helpful, even if somewhat artificial, to compare the resolution performance of a pixelated detector with that of a continuous-slab gamma camera. The resolution for a traditional Anger camera is typically characterized in terms of FWHM of the point spread function (PSF). For a 1D Gaussian-modeled PSF, 76% of events lie within a distance of ± 0.5 FWHM from the center. In 2D, 58% lie within a square, with each side equal to the

FWHM, around the center. Modern cameras using 9.5 mm-thick NaI (Tl) slabs, decoded with 76 mm diameter PMTs, attain 3.8 mm FWHM resolution over their UFOV, thus decoding 58% of events into the correct 3.8 mm square. Use of 51 mm diameter square PMTs with 9.0 mm-thick NaI (Tl) slab detectors improves the resolution to 3.1 mm FWHM over the UFOV [16], thus decoding 58% of events to the correct 3.1 mm square. In these continuous slab detectors, several cm (about half of the PMT diameter) of dead space is unusable at all edges. For our two-module detector with black and white modules and the initial PMT configuration, an average of 63% events are decoded to the correct pixel. Thus, 63% of incident events (over the entire area, including edges) are correctly decoded to lie within a 2.75×2.75 mm square. In that sense, our detector's average resolution is slightly better than a hypothetical continuous detector with an average 2.75 mm FWHM resolution and no dead space. This does not mean that the two types of detectors are equivalent; a pixelated detector has intrinsically finite linear sampling. However, there is an equivalence in this positioning metric, which serves to illustrate the effectiveness of detector pixelation in improving ISR and extending the usable fraction of the detector area. For the black module with all HQE PMTs, the fraction of correctly assigned pixels goes up to 72%.

Of course, the fraction of correctly decoded events in a pixelated detector can be greatly increased with finer spatial sampling of the scintillation light by effectively reducing the size of the light collection device relative to the pixel. For example, in the small animal SPECT imager [9] and PEM/PET system [10], mentioned in the introduction, 2 mm NaI detector pixels are successfully resolved by using arrays of 52 mm PSPMTs. The multi-anode Hamamatsu H8500 PSPMTs (Hamamatsu H8500 Multianode Photomultiplier Tube, Hamamatsu Photonics K.K., Shizuoka Pref., Japan) used in these systems contain an 8×8 array of anodes; thus, the sampling of light is performed at a significantly higher spatial frequency than for 51 mm single-anode PMTs described in the present work. However, after decades of development, PSPMTs such as H8500 are still a factor of ten more expensive than the single-anode PMTs described in this work.

Other approaches to finer light sampling, such as using avalanche photodiodes (APDs), position-sensitive APDs, and silicon photomultipliers, are well described by Madsen [2]. These methods, while yielding excellent resolution performance and low magnetic field susceptibility, also carry with substantial cost increases. Moving away from scintillation detectors, the two semiconductor-based cardiac SPECT systems [4], [5], mentioned in the introduction are based on 2.5 mm CZT pixels. Again, after decades of development, economic and materials issues of CZT detectors still limit their widespread use.

Perhaps more so in today's climate than ever, economic considerations must be taken into account along with technical merits. The tools have to be carefully chosen for the needs of the applications, with different technical, environmental, and economic constraints present in clinical research, small-animal imaging, and routine clinical nuclear medicine. In general, for clinical systems, detector cost increases lead to reduced total detector size. Because sensitivity is the predominant limitation of SPECT, we believe that the effective detector area is the system parameter that should not be compromised in clinical SPECT. This is especially true in the light of recent concerns about national radiation exposure [17]. The simultaneous pressures to reduce costs and patient radiation exposure are especially high for a widely used modality like cardiac SPECT, which accounts for nearly 60% of all nuclear medicine imaging in the U.S. [18]. It is for these reasons that we believe that the Anger camera and comparable systems are still the mainstay of nuclear medicine.

VI. Conclusion

Quantitative analysis has shown that the performance of our modular test detector yields very good event positioning for use in a gamma camera. This performance can be achieved over a curved detector arc that can be extended for complete body coverage, leading to significant advantage in terms of system geometric efficiency. Modular pixelated NaI(Tl) detectors used in conjunction with medium size conventional PMTs can achieve much of the ISR advantage of pixelation and meet the requirement for large-area curved detectors, while keeping costs contained. We therefore conclude that the use of pixelated NaI(Tl) in combination with relatively large single-anode PMTs, as described in this work, is a valuable option for meeting the demands of an important subset of clinical SPECT applications for the foreseeable future.

Acknowledgments

The authors would like to acknowledge D. Herr and L. Wilson at Saint-Gobain Crystals, for their role in the development of the detector, as well as G. Wong at M. Anderson and J. Karp at UPenn for leading them to the concept of using pixelation/PMT combination, and J. Strologas at Rush for reviewing the manuscript.

This work was supported the National Institutes of Health under Grant HL081414 and the Department of Energy under Grant DE-SC0002138.

Appendix

Detector Metrics by Region

The values of the detector metrics for seven module regions are given in Table III and Table IV. The module regions are defined in Fig. 17.

References

1. Anger HO. Scintillation cameral. Rev Sci Nucl Instrum. Jan.1958 29, no. 1:27–33.
2. Madsen MT. Recent advances in SPECT imaging. J Nucl Med. Apr.2007 48, no. 4:661–673. [PubMed: 17401106]
3. Chang W, Ordonez CE, Liang H, Li Y, Liu J. C-SPECT-a clinical cardiac SPECT/Tct platform: Design concepts and performance potential. IEEE Trans Nucl Sci. Oct.2009 56, no. 5:2659–2671. [PubMed: 23885129]
4. Esteves FP, et al. Novel solid-state-detector dedicated cardiac camera for fast myocardial perfusion imaging: Multicenter comparison with standard dual detector cameras. J Nucl Cardiol. 2009; 16, no. 6:927–934. [PubMed: 19688410]
5. CardiArc, Canton, MI.
6. Casey ME, Nutt R. A multicrystal two dimensional BGO detector system for positron emission tomography. IEEE Trans Nucl Sci. Feb.1986 33, no. 1:460–463.
7. Wong WH, Uribe J, Hicks K, Zambelli M. A 2-dimensional detector decoding study on BGO arrays with quadrant sharing photo-multipliers. IEEE Trans Nucl Sci. Aug.1994 41, no. 4:1453–1457.
8. Wojcik R, Majewski S, Kross B, Steinbach D, Weisen-berger AG. High spatial resolution gamma imaging detector based on a 5 diameter R3292 Hamamatsu PSPMT. IEEE Trans Nucl Sci. Jun.1998 45, no. 3:487–491.
9. Austin DW, Paulus MJ, Gleason SS, Mintzer RA, Siegel SB, Figueroa SD, Hoffman TJ, Wall JS. Design and performance of a new SPECT detector for multimodality small animal imaging platforms. Nucl Sci Symp Conf Rec. Oct.2006 2006:3008–3011.
10. Raylman RR, et al. The positron emission mammography/tomography breast imaging and biopsy system (PEM/PET): Design, construction and phantom-based measurements. Phys Med Biol. Feb. 2008 53, no. 3:637–653. [PubMed: 18199907]

11. Surti S, Karp JS, Muehllehner G. Evaluation of pixelated NaI(Tl) detectors for PET. *IEEE Trans Nucl Sci.* Feb.2003 50, no. 5:24–31.
12. Dahlbom M, Hoffman EJ. An evaluation of a two-dimensional array detector for high resolution PET. *IEEE Trans Med Imag.* Dec.1988 7, no. 4:264–272.
13. Wong WH, Hu G, Zhang N, Uribe J, Wang J, Li H, Lu W, Baghaei H, Yokoyama S. Front end electronics for a variable field PET camera using the PMT-quadrant-sharing detector array design. *IEEE Trans Nucl Sci.* Jun.1997 44, no. 3:1291–1296.
14. Gray RM, Makovski A. Maximum a posteriori estimation of position in scintillation cameras. *IEEE Trans Nucl Sci.* Feb.1976 23, no. 1:849–852.
15. Barrett HH, Hunter WCJ, Miller BW, Moore SK, Chen Y, Furenlid LR. Maximum-likelihood methods for processing signals from gamma-ray detectors. *IEEE Trans Nucl Sci.* Jun.2009 56, no. 3:725–852. [PubMed: 20107527]
16. Kimura K, et al. A new apparatus for brain imaging: Four head rotating gamma camera single-photon emission computed tomograph. *J Nucl Med.* May.1990 31, no. 5:603–609. [PubMed: 2341896]
17. Heller GV, et al. Importance of consideration of radiation doses from cardiac imaging procedures and risks of cancer. *J Nucl Cardiol.* 2010; 17, no. 1:1–3. [PubMed: 20066526]
18. 2008 Nuclear Medicine Market Summary Report. IMV Medical Information Division, Inc.; 2008. TMG2422031

Fig. 1. C-SPECT design concepts. (a) Patient imaging arrangement. (b) Inverse fan-beam collimation with factor of ~ 2 minification allows large number of simultaneous non-overlapping views. (c) A large-area curved detector system comprised of NaI(Tl) modules decoded with medium-size single-anode PMTs. Detector arc length is 120 cm.

Fig. 2.

(a) Top view photograph of the test modules. (b) Angled cutaway rendering of the modules. Plane of the cut is shown in (a). All diffuse reflecting surfaces are shown as white.

Fig. 3. 12-PMT test detector arrangement. (a) Cutaway side view. (b) Top view. The thick black lines indicate the black edge treatment applied on the side of the glass window of one of the modules.

Fig. 4.

Four PMT configurations used to evaluate the impact of HQE PMTs. Shaded circles indicate HQE PMTs. Only the 7 PMTs coupled to the black module and at the module joint were varied. (a) H0 configuration, baseline with all SQE PMTs. (b) H2, two HQE PMTs at the joint between the modules, configuration also used for module comparison. (c) H4, HQE PMTs at both transaxial edges of the black module. (d) H7, all HQE PMTs on black module.

Fig. 5.
Diagram of the DAQ system.

Fig. 6.

Beam scan set-up. A 0.7 mm circular hole is used to collimate a 140 keV source. (a) Set-up for 7.5° incidence. For all pixels, beam centerline crosses pixel centerline at pixel center. Presented beam shape represents 95% of beam's photons (simulated). (b) Set-up for measurements at normal incidence.

Fig. 7. Beam scan decoding map of the detector. The grayscale limits used throughout the paper are the min and max data values.

Fig. 8. R-error histogram. Histogram actually continues, with small number of occurrences, out to R-error of 61 (max possible value), but is cutoff for presentation.

Fig. 9.
Plot of R-error vs. beam-scan count rate for 2 different R-error cutoffs and for all events.

Fig. 10. Decoding error maps. (a) R error. (b) X error. (c) Y error.

Fig. 11.

(a) Pulse height resolution map. (b) Pulse height map.

Fig. 12.
Map of probability of event acceptance.

Fig. 13. Decoding maps for 4 different PMT combinations for the black module. (a) No HQE. (b) 2 HQE at module joint. (c) 4 HQE on corners. (d) 7 HQE. The same grayscale applies to all 4 maps.

Fig. 14. R-error maps for the above 4 PMT combinations. The same grayscale applies to all 4 maps.

Fig. 15. Pulse height resolution maps for the above 4 PMT combinations. The same grayscale applies to all 4 maps.

Fig. 16.
Energy spectra obtained from four pixels in the 7 SQE configuration at normal incidence.

Fig. 17.
Definition of module regions and relative module/lightguide/PMT positions.

Table I

Figures of Merit for the Black and White Modules

	$\overline{R_{er}}$	$\overline{X_{er}}$	$\overline{Y_{er}}$	f_c	f_{mm}	E/E, %	f_a
Black	0.41	0.27	0.20	0.66	0.97	10.2	0.88
White	0.52	0.33	0.25	0.60	0.95	10.0	0.88

Table II
Detector Metrics for Different PMT Combination (Indicated by Number of HQE PMTs)

	$\overline{R_{er}}$	$\overline{[X_{er}]}$	$\overline{[Y_{er}]}$	f_c	f_{nm}	E/E, %	f_d
H0	0.45	0.29	0.22	0.64	0.96	10.5	0.87
H2	0.42	0.26	0.21	0.67	0.96	10.2	0.87
H4	0.39	0.24	0.20	0.70	0.96	9.8	0.87
H7	0.36	0.23	0.18	0.72	0.96	9.1	0.84

Table III
Regional Detector Metrics for Black (B) and White (W) Modules

	$\overline{R_{er}}$	$\overline{X_{er}}$	$\overline{Y_{er}}$	f_c	f_{mm}	E/E, %	f_a
1-B	0.51	0.35	0.23	0.59	0.96	11.3	0.87
2-B	0.39	0.23	0.20	0.69	0.97	10.3	0.88
3-B	0.47	0.29	0.23	0.61	0.97	9.7	0.88
4-B	0.33	0.20	0.17	0.73	0.97	9.5	0.89
5-B	0.47	0.35	0.18	0.62	0.96	10.2	0.88
6-B	0.37	0.24	0.17	0.70	0.97	10.5	0.88
7-B	0.36	0.24	0.16	0.70	0.97	10.7	0.88
1-W	0.70	0.55	0.24	0.45	0.90	9.7	0.87
2-W	0.34	0.20	0.19	0.72	0.97	9.7	0.88
3-W	0.48	0.29	0.25	0.59	0.97	9.6	0.88
4-W	0.44	0.27	0.22	0.66	0.96	10.1	0.88
5-W	0.89	0.70	0.31	0.40	0.84	10.6	0.87
6-W	0.49	0.25	0.31	0.60	0.96	10.2	0.88
7-W	0.46	0.27	0.25	0.64	0.96	10.3	0.88

Table IV
Regional Detector Metrics for Different PMT Ccombinations (Indicated by Number of HQE PMTs), Black Module

	$\overline{R_{er}}$	$\overline{X_{er}}$	$\overline{Y_{er}}$	f_e	f_{nm}	E/E, %	f_d
1-H0	0.54	0.38	0.26	0.57	0.95	11.4	0.87
2-H0	0.42	0.24	0.23	0.67	0.96	10.3	0.88
3-H0	0.54	0.33	0.27	0.56	0.96	9.8	0.87
4-H0	0.38	0.22	0.22	0.70	0.96	10.2	0.87
5-H0	0.50	0.35	0.22	0.60	0.96	11.1	0.86
6-H0	0.40	0.26	0.19	0.67	0.97	10.7	0.88
7-H0	0.38	0.25	0.17	0.69	0.97	10.8	0.87
1-H2	0.53	0.36	0.25	0.59	0.95	11.3	0.87
2-H2	0.40	0.23	0.23	0.69	0.96	10.2	0.88
3-H2	0.48	0.28	0.25	0.61	0.96	9.7	0.87
4-H2	0.35	0.20	0.20	0.74	0.96	9.6	0.87
5-H2	0.47	0.33	0.20	0.62	0.96	10.1	0.86
6-H2	0.37	0.23	0.18	0.70	0.97	10.4	0.88
7-H2	0.36	0.23	0.17	0.71	0.97	10.6	0.87
1-H4	0.51	0.36	0.23	0.61	0.95	10.1	0.87
2-H4	0.35	0.20	0.20	0.73	0.96	9.5	0.88
3-H4	0.42	0.23	0.24	0.66	0.96	9.4	0.87
4-H4	0.34	0.20	0.19	0.74	0.96	9.5	0.87
5-H4	0.49	0.36	0.20	0.61	0.95	10.0	0.86
6-H4	0.34	0.21	0.17	0.73	0.97	9.9	0.88
7-H4	0.32	0.20	0.16	0.74	0.97	10.2	0.88
1-H7	0.45	0.30	0.21	0.65	0.95	9.7	0.84
2-H7	0.33	0.19	0.18	0.76	0.96	8.7	0.85
3-H7	0.41	0.25	0.21	0.68	0.96	8.5	0.84
4-H7	0.31	0.18	0.18	0.77	0.96	8.8	0.84
5-H7	0.46	0.34	0.18	0.63	0.96	9.7	0.83

	$\overline{R_{er}}$	$\overline{ X_{er} }$	$\overline{ Y_{er} }$	f_c	f_{nm}	$E/E, \%$	f_d
6-H7	0.32	0.20	0.15	0.76	0.97	9.1	0.86
7-H7	0.31	0.20	0.16	0.75	0.97	9.5	0.83