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The System Design, Engineering Architecture, and Preliminary Results of a Lower-Cost High-Sensitivity High-Resolution Positron Emission Mammography Camera

Yuxuan Zhang [Member, IEEE], Rocio A. Ramirez, Hongdi Li [Member, IEEE], Shitao Liu [Member, IEEE], Shaohui An, Chao Wang, Hossain Baghaei [Member, IEEE], and Wai-Hoi Wong [Member, IEEE]

M. D. Anderson Cancer Center, University of Texas, Houston, TX 77030 USA

Yuxuan Zhang: yuxuan.zhang@mdanderson.org

Abstract

A lower-cost high-sensitivity high-resolution positron emission mammography (PEM) camera is developed. It consists of two detector modules with the planar detector bank of $20 \times 12 \text{ cm}^2$. Each bank has 60 low-cost PMT-Quadrant-Sharing (PQS) LYSO blocks arranged in a 10×6 array with two types of geometries. One is the symmetric $19.36 \times 19.36 \text{ mm}^2$ block made of $1.5 \times 1.5 \times 10 \text{ mm}^3$ crystals in a 12×12 array. The other is the $19.36 \times 26.05 \text{ mm}^2$ asymmetric block made of $1.5 \times 1.9 \times 10 \text{ mm}^3$ crystals in 12×13 array. One row (10) of the elongated blocks are used along one side of the bank to reclaim the half empty PMT photocathode in the regular PQS design to reduce the dead area at the edge of the module. The bank has a high overall crystal packing fraction of 88%, which results in a very high sensitivity. Mechanical design and electronics have been developed for low-cost, compactness, and stability purposes. Each module has four Anger-HYPER decoding electronics that can handle a count-rate of 3 Mcps for single events. A simple two-module coincidence board with a hardware delay window for random coincidences has been developed with an adjustable window of 6 to 15 ns. Some of the performance parameters have been studied by preliminary tests and Monte Carlo simulations, including the crystal decoding map and the 17% energy resolution of the detectors, the point source sensitivity of 11.5% with 50 mm bank-to-bank distance, the 1.2 mm-spatial resolutions, 42 kcps peak Noise Equivalent Count Rate at 7.0-mCi total activity in human body, and the resolution phantom images. Those results show that the design goal of building a lower-cost, high-sensitivity, high-resolution PEM detector is achieved.

Index Terms

Nuclear imaging; nuclear medical applications; PET instrumentation

I. Introduction

Whole-body pet camera could be used for breast cancer diagnosis but has some shortcomings. First, there are lots of counts coming from the body rather than the breast which become the background noise and thus reduce the SNR. Second, the signals from breast tumor need to penetrate the patient's body before hitting the detectors, during which it causes the loss of signals. Moreover, a whole-body PET camera is very expensive thus it is not a cost-effective

way for breast imaging alone. Therefore, the development of dedicated positron emission mammography (PEM) cameras is more practical and useful for breast cancer applications.

Currently, there are a number of investigators who are developing the dedicated breast PET or PEM systems. In order to achieve a high-resolution capability that is needed for small tumor detection in breast cancer, most of those PET/PEM systems are using expensive Position-Sensitive PMTs or APD devices, which make their systems either expensive, and/or have very limited detector areas that affect the sensitivity [1]–[4]. Using our low-cost PMT-quadrant-sharing (PQS) techniques [5], [6] and the Slab-Sandwich-Slice method [7], we can build large-area detector banks with regular round shape PMTs at considerably lower cost, and achieve very high resolutions at the same time.

Compared to whole-body PET cameras, PEM detectors are usually put at a close distance, within a few centimeters from the breast, when used for imaging; thus the detection angle for a PEM system is usually very large which leads to higher sensitivity compared to those whole-body PET systems. A complete detector ring therefore is not necessary in terms of the sensitivity and FOV. Here we introduced the PEM system (MD Anderson PEM, or MDA-PEM) based on such concept.

II. System Design and Methods

A. Detector Bank Design

In the MDA-PEM system, there are only two planar detector modules. Each detector bank consists of 60 blocks made of LYSO crystal arranged in a 10×6 array with a total detection area of about $20 \times 12 \text{ cm}^2$, as shown in Fig. 1. Two types of blocks are used in the banks. The first type has the symmetric geometry which consists of 12×12 crystal arrays. The crystal size is $1.54 \times 1.54 \times 10 \text{ mm}^3$ and the pitch size is 1.62 mm. The dimension of the whole block is $19.36 \times 19.36 \times 10 \text{ mm}^3$. The second type is the 10 blocks in the top row of the bank (see Fig. 1) which has a rectangle shape. These blocks are extended in one direction and consists of 13×12 crystals with the dimension of $1.54 \times 1.93 \times 10 \text{ mm}^3$. The overall dimension of these blocks is $26.05 \times 19.36 \times 10 \text{ mm}^3$. From Fig. 1, we can see that the elongated blocks could cover almost all of the half empty PMT windows on the top edge of the detector bank that existed in PQS technique. Therefore, it will be more cost-effective since we can make a larger size detector panel with the same amount of PMTs. Moreover, it is more important for the PEM applications because we can reduce the dead area on the edge of the module which will be put against the patient's body during the data acquisition so that the chest wall region could be examined.

In our PQS design, all crystals are precisely cut and glued together. The gap between each crystal is minimized to 0.08 mm which is about the thickness of the reflector film (3M Enhanced Specular Reflector) used between crystals. Therefore the overall packing fraction of the whole detector bank is as high as 88%, which will help to achieve a very high sensitivity of the PEM system.

B. PMTs, Dividers and Electronics

There are 77 19-mm round PMTs (Photonis XP1912) in each module. We designed a new high-voltage divider that is soldered directly to each PMT. Compared to the socket type divider provided by the PMT manufacturer and used in our other system [8], the new divider's advantages are listed as following: 1) more reliable because no noise arises from the bad contact of a loose socket; 2) open design for better heat dispersion; 3) smaller dimensions which reduce the height of the module.

Front-end electronics are designed according to the faster decay time and higher light output of LYSO crystal. Detectors in each bank are grouped as four zones electronically. Each zone has its own Anger decoding electronics along with the high-yield-pileup-event-recovery (HYPER) [9], [10] technology that could handle the single event rate up to the 3 Mcps level. A new coincidence board with the hardware delay window for random coincidence events is developed. The coincidence window width can be adjusted from 6 to 15 ns. Since there are only two modules in the PEM system, the coincidence electronics is much simpler than that of a full PET system. The electronics diagram of the MDA-PEM signal processing chain is shown in Fig. 2.

C. PEM Module Design

The engineering of the modules is re-designed according to PEM applications based on the module design for a human PET system [8] with several major modifications and improvements. The inner structure of the module is shown in Fig. 3. The module box is made of aluminium alloy for a strong support and better cooling effect. Front-end electronics including the high-voltage dividers and the Anger boards are mounted inside the module box. The aluminium box could provide a good electromagnetic shield for the front-end electronics. The thickness of one of the sidewalls which will be put against the patient body is reduced in order to minimize the dead area for the detection. The distance from the crystal bank edge to the outer surface of the module is 7.9 mm in the current design and it could be further reduced in the future.

The crystal bed is made of black plastic material to reduce the absorption of gamma rays and provide light shielding. All detector blocks are glued together as a solid detector bank and then assembled inside the crystal bed with high mechanical accuracy. PMTs along with dividers are then coupled to the detectors with optically transparent material.

The total power consumption of each module is about 35 W, where about 22 W come from the dividers. Most of the energy will eventually convert to heat which leads to the increasing of the internal temperature of the modules. High temperatures and temperature variations will introduce noise and instabilities to the electronics, such as the dc drifts or PMT gain changes. Thus a low and stable temperature is important to ensure the stability of the whole system's performance. Semiconductor Peltier cooler devices are used in our PEM modules for cooling purposes. Three Peltier devices are attached to three sidewalls in each module. Heat sinks with fans are mounted on both inner and outer surfaces of the sidewalls helping the air circulation and heat dispersion. One sidewall that is close to the elongated blocks is left empty and will be put against the human body in clinical applications.

By adjusting the voltage applied to the Peltier devices, the cooling power could be changed in a wide range thus it is possible to keep a constant internal temperature when the environmental temperature changes. Compare to the compressed air cooling system we used before [8], using the Peltier cooler has several significant advantages. First, in order to use compressed air, there must be an air duct system inside the modules, which make the engineering design complicated and costly; moreover, because of the limited space inside the modules, the conductance of the air flow is relatively small, so the cooling efficiency is very poor and results in a higher working temperature. Using Peltier cooler could eliminate the air ducts, therefore, simplify the engineering design and improve the cooling efficient. The modules can work in a cooler environment under the new design. Second, the air compressor which is necessary for the air circulation in the old design is very noisy, and it could also blow dust into the modules even with air filters, which will affect the electronics especially the HV devices in the long term. The new cooling design is an airtight design; it eliminates the noisy air compressor and keeps a clean and quiet internal environment for the whole system.

D. Possible Data Collection Modes

Our PEM detectors have the detection area big enough to image the whole breast in most cases. The configurations to collect data with two planar detectors can be similar but not limited to the conventional X-ray mammography, as shown in Fig. 4(a), where the two detector banks are parallel and coaxial to each other. The side with the elongated blocks is put against human body to reduce the dead space, which is 7.9 mm as we mentioned before. The distance between two detector banks could be changed according to the compression of the breast.

In Fig. 4(b), we tilt the two modules so that the detector banks are no longer parallel but at an oblique angle between the detector planes. In this configuration, we can reduce the dead space caused by the module sidewall thickness so the detectors are closer to the human body than that in the conventional mammography mode. The area of the breast that can be imaged in each mode is shown in Fig. 4(a) and Fig. 4(b) as the shaded areas of the breasts indicate. Obviously the oblique mode covers more regions. In the configuration of Fig. 4(c), the axilla area could also be imaged which is very important for breast cancer diagnosis since in a lot of cases the cancer also spreads to the axillary lymph nodes when a patient has breast cancer. Fig. 4(d) is the arbitrary mode that the two detector banks are neither parallel nor coaxial, this mode might be necessary for some special cases.

The spatial sampling in mode **c** is the same as that in mode **a**, however, the breast could be compressed in mode **a** but not in mode **c**. And there will be more scattering for the gamma rays in mode **c** that may affect the image quality. The spatial samplings in mode **b** and **d** are different from mode **a**, which may affect the spatial resolutions. Moreover, conventional PET or PEM algorithms based on sinogram might not be appropriate in those cases because there is no full angular sampling in PEM imaging. Thus the dedicated PEM data binning and reconstruction algorithms are needed in order to get good image results.

During the data collection of PEM, there will be high single rates in each module due to the activities from the human body that is out of the FOV. Therefore the random coincidence rate might be very high and thus affect the Noise Equivalent Count Rate (NECR) and random events correction. In order to get a higher NECR, the PEM modules should be kept away from those hot organs such as the heart or bladder. The best approach for a clinical scan might be a compromise between the FOV, resolution, NECR, and may also be dependent on many factors such as the patient size, radiation intensity, tumor location, etc. It could only be optimized by detailed phantom and patient studies.

III. Performance Results

Two PEM detector modules have been built for the new MDA-PEM system. Some of the performance parameters have been studied and provided here.

A. Module Assembly

Fig. 5 shows the internal and external structures of the modules. Fig. 5(a) shows the assembly of PMTs with dividers and signal cables. The crystal bank is already mounted inside the black plastic part of the crystal bed and cannot be seen in this picture. In Fig. 5(b), the Anger boards and the sidewalls together with the heat sinks and fans are assembled. Three Peltier cooling devices are mounted on three sides of the module under the heat sinks, except one side which will be put against the human body during the data collection. Fig. 5(c) shows one complete module with back cover and all signal, power and HV cables.

B. Decoding Map

Fig. 6 shows the 2D decoding map of one module with ^{68}Ga flood source, where all crystals are clearly separated. All blocks have been examined in the test bench before the bank was assembled. The average peak-to-valley value of the blocks is 2.4.

For each event, the PMT signals in the x and y directions (EX and EY) and the full energy E are digitized and recorded in the raw data. Later the event location is calculated by EX/E and EY/E . From the 2D decoding map, we could extract the crystal lookup table and the individual crystal spectrum for further data analysis.

Fig. 7 is a typical positron source energy spectrum from one crystal in the blocks. Based on the flood source measurement, the average energy resolutions of all crystals in the two detector modules are $17.3\% \pm 3.5\%$ and $17.6\% \pm 3.6\%$, respectively.

C. Module Temperature

The working temperature inside the modules is around 30°C when the environmental temperature is 24°C . The temperature difference (6°C) is significantly lower than the compressed air cooling design which is typically 10 to 12°C . The temperature fluctuation is less than 0.5°C when environmental temperature does not have significant change. In the current design, the cooling system does not have the closed-loop feedback control for the cooling power, even though the temperature variation ($< 0.5^\circ\text{C}$) is acceptable and can be achieved if the PEM system is operated in an air-conditioned environment. Moreover, it is not difficult to upgrade the cooling control with a closed-loop feedback to maintain a constant working temperature if this system is operated in the environment with notable ambient temperature variations.

D. Sensitivity

The sensitivity of a PEM system is determined by the crystal material, dimensions, detector bank areas and distance from the source, etc. Since the distance between two PEM modules is changeable in different working conditions, the PEM system could have different sensitivities. We estimated the sensitivity of our PEM system by Monte Carlo simulations as the function of the distance between two parallel and coaxial banks with a point source located at the center of the FOV. The simulation is done by the dedicated nuclear medical software for Emission Tomography, GATE/Geant4 [11], [12], with the energy window of 400 to 650 keV . The sensitivity changes from 11.5% to 6.3% when the bank-to-bank distance changes from 50 to 100 mm , as shown in Fig. 8. Obviously, the sensitivity of the MDA-PEM is much higher than any other whole-body PET cameras.

E. Spatial Resolution

The spatial resolutions are measured by ^{22}Na point source and the results are shown in Table I. The point source is placed at the central plane between two modules and has a 1-cm transaxial offset from the center. Resolutions are tested with three bank-to-bank distances. The configuration of the coordinates system is shown in Fig. 9. Full 3D data are collected and binned into 2D sinograms in XZ plane for reconstruction by 2D FBP algorithm. The FWHM resolutions are derived from the reconstructions. From the results in Table I, we can see that the best resolution of our PEM system is about 1.2 mm , which is much higher than any other human PET cameras.

F. Count Rate Performance

The count rate performance of the MDA-PEM is determined by the time response parameters of the detectors and electronic circuits along the signal processing chain shown in Fig. 2. For

single events detection, the front-end electronics of each zone has a nonparalyzable deadtime of 120 ns; and the two multiplexers have a nonparalyzable deadtime of 50 ns. The single events from the two detector modules will then enter the coincidence board. The two coincidence logic circuits with 7.5 ns time windows are used to process the prompt and delayed coincidences events respectively. Two types of coincidences will then merge together and be stored in the FIFO buffer before being transferred to the PC computer through the PCI data acquisition board. The coincidence buffer has a 100-ns paralyzable dead-time. A raw energy cut threshold of 300 keV is set for singles in each zone, and a second energy cut window of 400 to 650 keV is applied to the coincidences during the data processing procedure.

The count rate performance is studied by Monte Carlo simulations. The phantom used in this study consists of two parts, one 70-cm-long and 20-cm-diameter big cylinder simulating the human torso, placed on the side of the detector banks. The other is a 10-cm-long and 10-cm-diameter small cylinder placed between the two banks simulating the breast. The distance between the two banks is 10 cm. The geometry of the phantom and the two detector banks is shown in Fig. 10. There is no shielding between the phantom and the detectors. Activity is uniformly distributed inside the phantom. The coincidence count rate curves as the function of the total activities and activity concentrations are shown in Fig. 11 and include the true coincidences, scattered coincidences, random coincidences and the NECR. From the simulation results, the peak NECR of the MDA-PEM is around 42 kcps with 7.0 mCi total activity.

Since there is no shielding in the detector modules, the single rates will be very high because the gamma rays coming from the radioactivity outside the FOV can hit the detectors, which may result in a high random coincidence rate. The coincidences from the prompt window and the delayed window are combined and outputted through one FIFO buffer which has limited bandwidth as described in Fig. 2. If the total rate is too high, some events will be lost during the data transfer process. Here we studied the coincidence rates of the prompt window and the delayed window after the FIFO buffer to see whether they match with each other under different activity conditions. The rates of these two random coincidences as the function of activity are shown in Fig. 12, which has a good accordance between the two curves. The average difference between the two random rates is only 5% over the entire activity range. Obviously, using delayed window method we can still get good estimation on the random coincidence for the MDA-PEM system.

G. Phantom Test

A Micro-Deluxe phantom filled with ^{68}Ga isotope is used for this test. The activity is about 0.3 mCi and the raw count rate is about 160 kcps (including both prompts and delayed window random events, before energy window cut). Data acquisition time is 15 min. During the data collection process, both the phantom and the PEM modules are in static positions. Therefore the data acquired only cover a limited angle range. The data are then binned as conventional 3D sinogram and reconstructed by 3D OSEM algorithm with 15 subsets and 15 iterations. 12 cross sections with 5 mm interval are shown in Fig. 13. We can see that the sectors with 4.8, 4.0, and 3.2 mm rods are clearly distinguished but not the sectors with rods smaller than 2.4 mm. Since the positron energy from ^{68}Ga is much higher than that from ^{18}F (maximum positron energy 1.90 MeV for ^{68}Ga and 0.64 MeV for ^{18}F), which means a much longer positron range for ^{68}Ga , the image quality is seriously affected by the ^{68}Ga positron range. The image can be greatly improved if ^{18}F isotope is used. Moreover, since PEM detector banks are very close to each other, the LORs from the edge crystals have a significant DOI effect which also degrades the image quality. Therefore, dedicated PEM reconstruction algorithms with DOI corrections could improve the image quality and resolution further, which is under development in our lab.

IV. Conclusion

A lower-cost, high-sensitivity, high-resolution PEM camera is successfully developed with the PQS detector design. The small crystal size of 1.54 mm ensured the high-resolution capability. The large area detector banks result in the high sensitivity of 11.5% to 6.3% for point source with the bank-to-bank distance of 50 to 100 mm. The overall energy resolution is around 17% for both detector modules. Good detector-decoding maps are obtained and all crystals are clearly distinguished. Spatial resolution is measured with ^{22}Na point source with the best result around 1.2 mm. The simulation shows a very high NECR with the peak value of 42 kcps at 7.0 mCi total activity with the human phantom. The micro deluxe phantom images are obtained with ^{68}Ga source and could be further improved by ^{18}F source and better image reconstruction software.

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Fig. 1. Geometries and dimensions of the MDA-PEM detector bank (top) and two types of the blocks (bottom). Bottom-left: 12×12 symmetric block; Bottom-right: 13×12 asymmetric elongated block.

Fig. 2.
Electronics diagram of the signal processing chain.

Fig. 3.
PEM module inner structure.

Fig. 4. Some of the suggested data collecting modes. (a) Conventional X-ray mammography mode. (b) Oblique mode. (c) Axilla mode. (d) Arbitrary mode.

Fig. 5. PEM detector module assembly. (a) Detector bed with PMTs, HV dividers and signal cables. (b) Sidewalls with heat sinks and fans, Anger board on the top. (c) Complete detector module with back cover and cables for signal, data, and power.

Fig. 6.
Part of the 2D decoding map from one module with flood source.

Fig. 7.
A typical positron source energy spectrum from one crystal chosen from the detector bank.

Fig. 8.
Point source sensitivity as the function of the distance between two modules.

Fig. 9.
Coordinates configuration for the spatial resolutions measurement.

Fig. 10.
Phantom geometry for count rate simulations.

Fig. 11.
Coincidences count rates as the function of total activities and activity concentrations.

Fig. 12.
Comparison of the random coincidence rates coming from the prompt window and the delayed window.

Fig. 13. Reconstructed micro-deluxe phantom with 5 mm interval between slices, with data from ^{68}Ga isotope. No attenuation, scattered and random coincidence corrections are applied.

TABLE I

FWHM Resolutions

D (mm)	R_x (mm)	R_y (mm)	R_z (mm)
50	1.43	2.40	4.00
70	1.19	2.01	4.10
90	1.29	2.21	4.82

D: Banks-to-bank distance

²²Na point source located at the center plane between two detector planes and with 1-cm offset from the center of the plane.