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Arrays of Segmented, Tapered Light Guides for Use with Large, Planar Scintillation Detectors

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Abstract

Metabolic imaging techniques can potentially improve detection and diagnosis of cancer in women with radiodense and/or fibrocystic breasts. Our group has previously developed a high-resolution positron emission tomography imaging and biopsy device (PEM-PET) to detect and guide the biopsy of suspicious breast lesions. Initial testing revealed that the imaging field-of-view (FOV) of the scanner was smaller than the physical size of the detector's active area, which could hinder sampling of breast areas close to the chest wall. The purpose of this work was to utilize segmented, tapered light guides for optically coupling the scintillator arrays to arrays of position-sensitive photomultipliers to increase both the active FOV and identification of individual scintillator elements. Testing of the new system revealed that the optics of these structures made it possible to discern detector elements from the complete active area of the detector face. In the previous system the top and bottom rows and left and right columns were not identifiable. Additionally, use of the new light guides increased the contrast of individual detector elements by up to 129%. Improved element identification led to a spatial resolution increase by approximately 12%. Due to attenuation of light in the light guides the detector energy resolution decreased from 18.5% to 19.1%. Overall, these improvements should increase the field-of-view and spatial resolution of the dedicated breast-PET system.

Index Terms

Nuclear Medicine; Detectors; Breast Cancer

I. Introduction

Difficulty in detection of breast cancer in the subset of women with radio-dense and/or fibrocystic breasts using standard mammography (1–3) was a significant factor in spurring the development of specialized breast imaging systems. Many of the early breast scanners utilized Nuclear Medicine methods (single photon and positron emission), due to the potential advantages these techniques offer for cancer detection in this group of women. In addition to the development of gamma cameras using single photon-emitting radionuclides for dedicated breast imaging (4–9), scanners utilizing positron-emitting radionuclides were developed. The first of these systems, known as positron emission mammography (PEM), was proposed in 1994 by Thompson *et al.* (10). Since this initial work, there have been a number of proposed and constructed dedicated positron emission imagers (11–17).

In addition to these efforts, the Nuclear Medicine Instrumentation group at West Virginia University in collaboration with the Thomas Jefferson National Accelerator Facility (Newport News, VA) developed a PET imaging system to detect and guide the biopsy of positron emission radiotracer-avid breast lesions (18). Called PEM-PET, this first generation scanner consisted of two pairs of detector heads, each based on a 96×72 array of $2 \text{ mm} \times 2 \text{ mm} \times 15 \text{ mm}$ LYSO scintillator elements, pitch = 2.1 mm (detector size = $20.2 \text{ cm} \times 15.1 \text{ cm}$). The individual elements were polished on all sides and optically separated from adjacent elements by 0.1 mm-thick strips of white reflective polyester material (Saint Gobain Crystals, Newberry OH). The scintillator array and housings were designed to add minimal distance between the scintillator array and the top of the units to maximize sampling of areas close to the chest wall. The front surface of each array was covered with a white reflective coating and a thin carbon fiber reinforced composite material. A 4 mm-thick glass window covered the other surface of the array; this end was coupled to a 4×3 array of Hamamatsu model H8500 flat panel position-sensitive photomultiplier tubes (PSPMTs). This system was tested in an initial clinical trial in 2011 (19) and has since been undergoing an upgrade to include x-ray CT imaging capabilities.

A 2 mm-thick solid acrylic light guide was mounted between the scintillator array and PSPMT array to spread scintillation light among several elements of the PSPMTs. Distribution of scintillation light was required to permit accurate determination of photon interaction position in the scintillator array. The position of each photon interaction in the scintillator array was determined by calculation of the center-of-mass of the light distribution detected by the PSPMT array. This information was then used in conjunction with a previously calculated lookup table to convert the position of the event to the appropriate detector element number. Finally, the energy of each event was calculated using a pre-measured table that converts signal amplitude to energy.

The PSPMTs were mounted as close to the edges of the scintillator array as possible to collect light from these areas. Nevertheless, it was not possible to identify the top and bottom rows, and left and right columns of detector elements due to limitations in collection of light at the periphery of scintillator array. The front faces of the H8500 are made of glass with the cathode material deposited on its inner surface. At the edges of the windows, the glass, and thus the cathode curves where it meets the metal case of the PSPMT. Hence, the

trajectories of emitted photoelectrons and the propagation of the electron showers to the anodes becomes distorted, which can introduce ambiguities in localization signal origin. Additionally, light passing through the curved part of the window can be refracted, causing mis-positioning of light signals from the edges of the scintillator array. The effect of these interactions is that light from the two rows and columns at the edges of the array can become superimposed on the adjacent row or column, forming a single row or column. Hence, the scintillator array was effectively not 96×72 , but 94×70 , which meant that the full area of the detector (including the top edge adjacent to the chest wall of a patient) was not contributing usable imaging data.

II. Materials and Methods

A. Light guides

To address the incomplete collection of light at the edges of the scintillator array, a new method for optically coupling the scintillator to the PSPMTs was developed. Specifically, the solid, 2 mm-thick acrylic light guide was replaced by an array of tapered light guides. In this scheme, each of the twelve PSPMTs is individually coupled to the LYSO array via a tapered, pixelated 5.3 mm-thick light guide constructed in collaboration with Agile Technologies (Knoxville TN) (these devices are now commercially available). The light guides were constructed from an array of 16×16 individual slanted glass optical elements that increase in slant angle as the distance from the center of the light guide is increased (Figure 1). The light guide was designed to condense the light collected from the front face of the structure to an area that is slightly smaller than the active area of an H8500 PSPMT ($49 \text{ mm} \times 49 \text{ mm}$), while preserving positional signal integrity. Thus, the scintillation light does not pass through the outer edges of the H8500's entrance window. Note that the size of the front face of the light guide ($52.5 \text{ mm} \times 52.5 \text{ mm}$) is an integral number of detector elements (25×25 detector elements) and matches the outer dimensions of the H8500. Therefore, the seams of the light guide array coincided with edges of the detector element array, minimizing the likelihood that light from a detector element is split between adjacent light guides. Some multiplexing of the signal within the light guide was permitted. The ratio of detector element number to the number of light guide channels is 2.5:1 in the mid section, 1.6:1 along the edges, and 1:1 at the corners. The lower detector element to light guide channel ratio at the edges, and even lower at the corners was introduced to overcome the difficulty of getting good edge position information. The height of the light guide (5.3 mm) was kept as small as practical to minimize the physical size of the detector. Each light guide contains 256 individual elements, separated from its neighbor with a white polyester reflective coating. Figure 2 shows an individual light guide and a 4×3 array of light guides mounted on the scintillator array.

Using a positioning jig, the edges of the light guides were aligned with the edges of the scintillator array. This scheme is intended to increase the probability that light from detector elements at the edges of the LYSO array will be detected. Coupling of the light guides to the scintillation arrays and PSPMTs was performed with a semi-rigid, silicone-based gel that provided good light transport and mechanical stability. Note that tapered light guides have previously been used to optically coupling scintillator to PSPMTs (20–21). These structures,

however, consisted of monolithic pieces of optically transparent material and were used to couple individual, small scintillator arrays to PSPMTs to facilitate arrangement of the detector modules into rings. Fiber optic light guides were used in the MicroPET systems (22), but they were not tapered. Again, the reason for their use was to facilitate the arrangement of the individual detector modules in a ring geometry, unlike our application to large planar scintillation detectors.

B. Performance testing

To gauge the effect of the use of the tapered light guides, a $20\text{ cm} \times 15\text{ cm} \times 4\text{ cm}$ water-filled acrylic phantom, containing $60\text{ }\mu\text{Ci}$ of ^{18}F was placed midway between two of the detectors operating in coincidence. A 60 min data acquisition was performed. These coincidence data were used to create maps of the calculated positions of the scintillator detector elements (also known as crystal maps). The ability of the system to precisely identify the positions of the detector elements was evaluated by calculating the peak-to-valley contrast ratio (PVCR). PVCR is defined as the maximum amplitude of the signal from an area of a detector element (peak) minus the minimum amplitude of the adjacent area between peaks (valley) divided by the amplitude of the peak. PVCRs were measured using automated software that identified the peaks and valleys in a crystal map. The potential effect of improved detector element identification was assessed by measuring spatial resolution of the two generations of scanner. A point-like source of ^{18}F was scanned at different distances from the scanner center-of-rotation (COR). The data were reconstructed using the OSEM algorithm (25° acceptance angle, three subsets and ten iterations) (23). The full-width-at-half-maximum (FWHM) of the point-spread function was measured from the resulting images. Finally, positions of photopeaks measured from ROIs placed on the location of peaks in the crystal maps at three important locations (center of a PSPMT, seam between adjacent PSPMTs and a junction where four PSPMTs meet) were measured.

III. Results

Fig. 3 shows the crystal maps for the first- and second-generation detectors. Note that in the crystal map generated by the first generation detector only 94×70 of the 96×72 detector elements were discernable, whereas all of the elements are discernable in the map produced with the second-generation detector. Fig. 4 shows plots of scanner spatial resolution measured for generation 1 and 2 of the detectors, as well as a plot showing the resolution differences between the two generations. Fig. 5 displays plots of the global detector PVCR, and PVCRs at seams between PSPMTs and top row of the detector. The error bars show the standard deviation of the mean. For each region, an unpaired t-test was performed to assess if there were statistically significant changes in mean PVCRs between the first- and second-generation detectors. In all cases, the mean PVCRs for the second-generation detectors were significantly higher ($p < 0.01$, $\alpha = 0.05$) than the mean PVCRs measured for the first-generation scanner. Specifically, the mean PVCR for the entire detector increased by 50%. At the seams between rows of PSPMTs the PVCRs increased by 129%. For the top two rows of detector elements the PVCRs increased by 38%. Table I shows a comparison of the photopeak positions at areas of interest in the scintillator array (central area of a PSPMT,

seams between PSPMTs and junction of four PSPMTs). Finally, the system energy resolution of the first generation detector was 18.5% and 19.1% for the second detector.

IV. Discussion

The goal of this project was to improve the localization of photon interactions in our detector and extend its FOV. The accurate positioning of photon events is important for maximizing the spatial resolution of a scanner. Extending the sampling of areas close to the chest wall may permit better detection of suspicious lesions in this challenging area of the breast. The detector heads of our system were designed to have ~3 mm of housing material between the top of the scintillator array and the top edge of the detector housing. In the first-generation scanner, this distance was effectively increased by at least an additional 2.1 mm (the pitch of the detector elements) because it was not possible to obtain discernable signal from the top row of scintillator. This limitation was caused by the non-optimal optics at the edges of the H8500 PSPMT. We found that the use of segmented, tapered light guides to transport light from the edges of the scintillator array to active areas of the PSPMT face facilitated detection of light from the detector elements at the edges of the scintillator array. The crystal maps shown in Fig. 3 demonstrate the effect of the new optics on identification of detector elements. Specifically, the crystal map obtained with the second-generation detector contains all 96×72 elements, whereas the first-generation map contained 94×70 resolvable elements. Therefore, the effective area of the detector was increased by two rows and two columns of detector elements.

The bright rim around the crystal map from the first generation detector is produced by the optical superposition of two rows and columns caused by the complex photoelectron trajectories and refractive optics at the edges of the H8500 entrance window. Use of the new light guides addresses these phenomena by transmitting the light from the edge of the scintillation array to the active area of the H8500 where the entrance window is flat and there are minor optical distortions. Thus, the bright edges are resolved into two rows and columns.

In addition to transmission of light from the edges of the scintillator array to the active area of the PSPMTs produced by the tapering of the light guide, the segmented light guides act as collimators, confining the angle of the light distribution cone. This effect facilitates precise calculation of the center-of-mass of the light distribution by limiting the spread of light to a smaller number of anodes. Thus, the data used to perform the center-of-mass calculation has higher count densities, resulting in more precise position calculations. One of the advantages of more precise positioning of detector elements is slightly improved spatial resolution, as demonstrated by the results shown in Fig. 4. System spatial resolution was increased by approximately 12% by the use of the new light guides.

The increased PVCR values shown in the plots of Fig. 5 quantitatively demonstrate improvements in the accuracy of event positioning achieved via the use of the tapered light guides. The PVCR for the whole detector increased by 50%, which quantitatively confirms the visual impression that the second-generation crystal map appears “sharper” than the first generation map. The large improvement in contrast at the seams between PSPMT rows

(129%) is due to the increase in accuracy in placement of events in these optically challenged regions. Perhaps the most noteworthy increase is at the top rows of the detector, where the PVCR increased by 38%; indicating that not only did the new optical coupling method allow discrimination of the top row of detector elements it also improved the accuracy in localizing event position in this area. These changes should aid in imaging of the areas of the breast close to the chest wall. Note that we used PVCRs instead of the more common peak-to-valley ratios measured from intensity profiles drawn on the crystal maps to assess the accuracy of event positioning due to the inaccuracies inherent in drawing these profiles. Specifically, rows and columns in the crystal maps are often not straight due to small optical discontinuities, so it is difficult to consistently sample the peaks and valleys with a linear profile. The use of a two-dimensional, automated algorithm to identify peaks and adjacent valleys to calculate PVCRs removed this source of ambiguity from the analysis.

Finally, the results presented in Table I showing the position of the photopeaks at three representative areas of the detector further illustrate the effects of the light guides on event positioning. Photopeak position is representative of the amplitude of the amount of photons detected in these areas. Thus, the results for the first generation detector shows that there is a large disparity among the three different areas due to the effects of optical distortions in the PSPMTs described above. Specifically, as the light from the detector elements is distributed between two H8500s (at seams) and among four devices (at PSPMT junctions), the total amount of light collected is reduced compared the center of a PSPMT where light collection is most efficient. Use of the light guides has two effects. First, the amount of light transmitted to the central areas of the PSPMTs is reduced by 27% due to attenuation of light in the material used to construct the light guides. Second, the new light guides equalized the intensity of the light collected across the face of the detector. This effect results in relatively uniform light collection efficiency across the detector, unlike the first generation detector where there was a 56% variation between the maximum intensity (center of a PSPMT) and the minimum intensity (PSPMT junction). The slight overall loss of scintillation light intensity incurred by the introduction of the tapered light guides resulted in a slight loss of energy resolution (18.5% versus 19.1%).

V. Conclusions

In summary, two weaknesses in the original design of the large, flat panel scintillator detector constructed for use in a dedicated breast-PET scanner were addressed by implementing a new type of tapered light guide. Specifically, these devices extended the FOV of the detector by ~2 mm at the edges of the detector, facilitating the sampling of areas of the breast closer to the chest wall. The new light guides also enhanced the accuracy of event positioning, leading to improved spatial resolution, and increased the uniformity of light collection in the detector. The effect of these enhancements on overall scanner performance will be evaluated once the breast-PET/CT that incorporates the second-generation PET scanner is completed.

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Fig. 1.
Schematic drawing of a light guide. Units are in mm.

Fig. 2. Picture showing one of the tapered light guides (top). Picture showing the 4×3 array of tapered light guides mounted on the scintillator array (bottom).

Fig. 3. Crystal maps for the first generation detector that utilized a solid light guide (top) and for the second-generation detector that used the segmented, tapered light guides (bottom).

Fig. 4.

Plots of spatial resolution measured for the two generations of detectors as a function of position (a) and plots of percent differences in spatial resolution between the two generations of detectors as a function of source position (b).

Fig. 5.

Comparisons of PVCR for the two generations of detectors at three locations (a) global value, (b) seams between PSPMTs and (c) top row of the detector.

Table I

Positions of Photopeaks Measured at Three Locations

<u>Generation</u>	<u>Position of Photopeak (Channel Number)</u>		
	<u>Center</u>	<u>Seam</u>	<u>Junction</u>
First	123	84	54
Second	90	93	95