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Temperature dependent operation of PSAPD-based compact gamma camera for SPECT imaging

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Abstract

We investigated the dependence of image quality on the temperature of a position sensitive avalanche photodiode (PSAPD)-based small animal single photon emission computed tomography (SPECT) gamma camera with a CsI:Tl scintillator. Currently, nitrogen gas cooling is preferred to operate PSAPDs in order to minimize the dark current shot noise. Being able to operate a PSAPD at a relatively high temperature (e.g., 5 °C) would allow a more compact and simple cooling system for the PSAPD. In our investigation, the temperature of the PSAPD was controlled by varying the flow of cold nitrogen gas through the PSAPD module and varied from −40 °C to 20 °C. Three experiments were performed to demonstrate the performance variation over this temperature range. The point spread function (PSF) of the gamma camera was measured at various temperatures, showing variation of full-width-half-maximum (FWHM) of the PSF. In addition, a ^{99m}Tc-pertechnetate (140 keV) flood source was imaged and the visibility of the scintillator segmentation (16×16 array, 8 mm × 8 mm area, 400 μm pixel size) at different temperatures was evaluated. Comparison of image quality was made at −25 °C and 5 °C using a mouse heart phantom filled with an aqueous solution of ^{99m}Tc-pertechnetate and imaged using a 0.5 mm pinhole collimator made of tungsten. The reconstructed image quality of the mouse heart phantom at 5 °C degraded in comparison to the reconstructed image quality at −25 °C. However, the defect and structure of the mouse heart phantom were clearly observed, showing the feasibility of operating PSAPDs for SPECT imaging at 5 °C, a temperature that would not need the nitrogen cooling. All PSAPD evaluations were conducted with an applied bias voltage that allowed the highest gain at a given temperature.

Index Terms

Avalanche photodiodes; Biomedical nuclear imaging; SPECT; Scintillation detectors

I. Introduction

Arrays of photomultiplier tubes (PMTs) have been used to produce position and energy information of the incident radiation events ever since Anger's pioneering scintillation gamma camera for nuclear medicine applications [1],[2]. PMTs provide high signal gain and low electronic noise at relatively low cost and are a mature photon detector technology. However, their fragility and bulkiness stemming from the vacuum tube technology is a major disadvantage. Solid-state detectors have been suggested as replacements for PMTs, and extensive research and development has occurred in recent years. For example, an array of silicon avalanche photodiodes (APDs) coupled with a scintillator has been investigated, but this device suffers from a large number of readout channels and dead space between the APD array elements [3]–[5]. Cadmium zinc telluride (CZT) provides excellent energy resolution for scattered photon rejection, fine intrinsic spatial resolution, and a high signal-to-noise ratio (SNR) at room temperature, outperforming PMT-based detectors [6]–[8]. However, high manufacturing cost, limited supply and a large number of readout channels limit the widespread adoption of CZT-based gamma cameras.

Silicon-based position-sensitive avalanche photodiodes (PSAPDs) are attractive as photodetectors due to their intrinsic position sensing capability and low number of required readout channels (4 channels per PSAPD), good spatial resolution, high gain and lower cost compared with CZT technology [9]. In addition, PSAPDs are mechanically robust and geometrically compact, allowing the construction of small, portable radionuclide systems for small animal imaging or targeted human organ imaging such as heart or brain [10]–[12]. However, there are two major disadvantages to PSAPDs, shot noise induced by the capacitance from relatively large surface area of the PSAPD, and consequently, a performance sensitivity to operating temperature.

Dark current shot noise is the main source of temperature dependent noise in a PSAPD. Research on PSAPDs for positron emission tomography (PET) application has shown that a PSAPD can be operated at room temperature for detecting 511 keV photons [13],[14], but no such temperature dependent performance evaluation study has been performed for SPECT applications, which use lower energy photons (e.g., 140 keV for $^{99\text{m}}\text{Tc}$). Here, we evaluated the operating temperature dependence for a PSAPD by measuring the quality of flood images and by determining the full-width-half-maximum (FWHM) of the point spread functions (PSFs) at a range of temperature from $-40\text{ }^{\circ}\text{C}$ to $20\text{ }^{\circ}\text{C}$. The PSF is a measure of the image quality that is affected by both PSAPD noise characteristics and CsI:Tl scintillator response. At each temperature, the applied bias voltage was adjusted in order to avoid breakdown voltage of the PSAPD ($\sim 10\text{ V}$ below breakdown point) and maximum gain available. The reverse breakdown voltage of APDs increases with temperature, and a higher reverse bias voltage will increase the gain. However, the dark current also increases with reverse bias voltage, so at a given temperature, the applied bias voltage affects both gain and shot noise [16]. Then, we investigated the quality of two reconstructed images of a mouse heart phantom by acquiring tomographic data at the cold operating temperature ($-25\text{ }^{\circ}\text{C}$) as well as at $5\text{ }^{\circ}\text{C}$ to determine the feasibility of using PSAPDs at temperatures that do not require cooling by liquid nitrogen.

II. PSAPD

The PSAPD used in this work was developed and fabricated at Radiation Monitoring Devices (RMD), Inc. (Watertown, MA). It has an $8\text{ mm} \times 8\text{ mm}$ active area and is packaged on a ceramic substrate. Within the PSAPD, photo-induced electrons are amplified through an avalanche process and travel across a resistive layer toward four anode corners. Charge-sharing amongst the four anodes combined with Anger-logic gives the PSAPD a position

sensing capability [2]. As shown in Fig. 1, the photograph of the PSAPD shows a square surface with 4 anodes and a continuous CsI:Tl scintillator used for experiment. The energy and timing of the radiation events can be extracted from the signal using a common cathode. The PSAPD has an operating gain of approximately 1000 for a bias voltage of ~ -1600 VDC with ~ 200 electrons-rms noise at room temperature and a $\sim 60\%$ quantum efficiency (defined as generated electron-hole pairs/incident photons) in the 400 to 700 nm region [15]. The breakdown voltage of the PSAPD depends on the operating temperature. The negative bias voltage can be increased at higher temperature [16]. For example, the applied bias voltage for operation at -40°C was -1600V , -25°C was -1640V and for 5°C was -1710V , which were set $\sim 10\text{V}$ below the breakdown voltage ($\sim 10\text{V}$). These same voltages were applied for the flood image acquisitions, PSF measurement and mouse heart phantom study for each temperature. Electrical signals from the PSAPD were pre-amplified by Cremat-110 charge sensitive preamplifiers (Cremat, Inc. Watertown, MA) with 1.4 gain (V/pC) and then amplified and shaped by pulse-shaping amplifiers (Canberra 2020, Canberra Industries, Meriden, CT) with 250 ns pulse shaping times and 35 dB gains. The temperature of the PSAPD was controlled by adjusting the flow of nitrogen gas through the PSAPD containment vessel.

III. PSF Measurement Versus Temperature Variation

In order to quantify the temperature dependence of the PSAPD performance, we measured the width of the PSF at temperatures from -40°C to 20°C in 5°C increments with the bias voltage set $\sim 10\text{V}$ below the breakdown voltage, thereby achieving high gain. The PSF measurements at different temperatures demonstrate the effect of dark-current shot noise in the projection image, and are efficient methods to investigate the feasibility of using PSAPD at close-to-room temperature with a simple cooling for SPECT applications. A 2-mm thick continuous CsI:Tl crystal was used instead of the segmented crystal in order to take advantage of the high positioning capability of PSAPD independent from scintillator pixellation. The PSAPD and crystal combination was placed in a pinhole gamma camera configuration with a 0.5 mm tungsten knife-edge pinhole collimator (60 degree acceptance angle and 15 mm thickness), with a 33 mm spacing between the pinhole and the CsI:Tl scintillator surface coupled to the PSAPD. A 370 MBq $^{99\text{m}}\text{Tc}$ -pertechnetate source with a diameter of 5 mm was placed 0.5 m away from the pinhole, yielding a 1/15 minification of the source. Due to the large minification, the far-field source can be regarded as a point source at the image plane. A 15% energy window centered at 140 keV was applied and the pincushion distortion was corrected using a rotated diagonal readout algorithm [17]. The algorithm uses corresponding diagonal pairs to calculate a scintillation light centroid and reorient the image using a 45° rotation. The algorithm generates images with reduced pincushion distortion compared with standard Anger-logic due to more uniform charge sharing along the diagonal. Post-processing (bias subtraction) was applied to filter out non-thermal bias build up caused by problem with sample-and-hold board and reflected photons from the sidewall of the CsI:Tl scintillator. Fig. 2 shows the widths (mean value of 7 PSF FWHM measurements) and error bars (standard deviation of 7 PSF FWHM measurements) of the PSF FWHM against the measured temperature on the PSAPD after a Gaussian fit was applied to the PSF data. The measured FWHM of the PSF varies with temperature with small increase (slope = 0.0018 from a linear regression analysis to the data) as shown in Fig. 2.

IV. Uniformity Versus Temperature Variation

The uniformity variations of the PSAPD with temperature can be used to demonstrate the quality of image at different operating temperatures of the PSAPD using the mean-to-standard deviation ratio as a metric. We flooded the PSAPD that was optically coupled to

the 2-mm thick, 8 mm $\times$ 8 mm area, continuous CsI:Tl crystal, with a 1,480 MBq ^{99m}Tc -pertechnetate-filled cylindrical volume with 1-cm diameter and 1-cm height cylindrical volume within a syringe. The flood source was placed 30 cm away from the PSAPD. Flood images were obtained at different temperatures from -40°C to 20°C with a 5°C increment with a corresponding bias voltage (just below breakdown voltage) applied for each flood image acquisition. For each flood image, 500,000 counts were accumulated with a 7-second counting time. A square region of interest (ROI) having a dimension of 6 mm $\times$ 6 mm was defined inside the flood image, and we calculated the mean-to-standard deviation ratio of total counts inside the ROI. Fig. 3 shows the measured mean-to-standard deviation ratio (mean value of 5 measurements) plotted against the operating temperature. The error bars (standard deviation) were calculated from 5 measurements at each temperature setting. The standard deviation could be affected mostly by temperature dependent shot noise and small gain variation over the surface of the PSAPD. The result illustrates that the uniformity generally degrades as the temperature increases. The peak was found at -25°C .

V. Flood Image

Flood images obtained using a segmented scintillator at different temperatures provides another measure of the temperature-dependent noise properties of the PSAPD. The PSAPD was optically coupled to a 16 $\times$ 16 segmented CsI:Tl scintillator (8 mm $\times$ 8 mm area, 3 mm thick, 69% stopping power at 140 keV), placed inside a module, and cooled by nitrogen gas. The same 1,480 MBq ^{99m}Tc -pertechnetate flood source as in the uniformity experiment described above was used to obtain flood images at three temperatures (-40°C , -25°C , and 5°C) with corresponding bias voltages at -1600 V , -1640 V , and -1710 V respectively. Fig. 4 shows flood images taken at these temperatures. Each image has a highly nonlinear pincushion distortion inherent to Anger-logic calculations [9],[18]. Fig. 4(a) shows a flood image at -40°C and exhibits poor definition of the scintillator pixels both in the central area and at the edges. Fig. 4(b) shows a flood image taken at -25°C , qualitatively illustrating clear pixel separation. At 5°C , the dark current shot noise in PSAPD increases the noise level possibly surpassing SNR improvement by larger gain at higher temperature, generating a noisy flood image as shown in Fig. 4(c). We also quantified the flood images at these temperatures by measuring the mean values (and standard deviations) of the peak-to-valley ratios across 9 pixels located around the center of the scintillator. The mean values (and standard deviations) of the peak-to-valley ratio of the flood image were 5.34 (1.95) at -40°C and 6.28 (1.96) at -25°C . For the flood image at 5°C , the lowest mean peak-to-valley value of 4.44 (1.22) was observed as expected.

VI. Phantom Experiment Description

PSAPD-based SPECT data were obtained from the imaging setup shown in Fig. 5. A phantom simulating the left ventricular wall of a mouse heart (10.7 mm height, 4.9 mm diameter, and 0.85 mm wall thickness) and a cold defect (2 mm $\times$ 6 mm in size, 0.7 mm thick) was filled with a 370 MBq aqueous solution of ^{99m}Tc -pertechnetate aqueous solution. The mouse heart phantom was imaged using a 0.5 mm diameter knife-edge tungsten pinhole with a 60° acceptance angle. The ratio of distances (from the PSAPD to the pinhole/from the center of the object to the pinhole = 33 mm/55 mm) produced a 60% minified image of the phantom on the PSAPD that was coupled to a 2-mm thick continuous CsI:Tl scintillator. The mouse heart phantom was mounted on a rotation stage for tomographic data acquisition. A three-axis translation stage was used to control the location of the phantom. Two separate tomographic data sets were acquired, one at -25°C and the other at 5°C , each with $\pm 0.5^\circ\text{C}$ fluctuation during data acquisition. We acquired 34 projections at 6° increments covering 204° of the phantom. For each projection, 200,000 counts were accumulated, and the acquisition time varied from 2 to 8 minutes, due to radioactive decay of the source over the

course of the experiment, yielding an average count rate of 667 counts/sec and a sensitivity of 1.8×10^{-6} . The number of counts inside the 15% energy window centered at 140 keV, i.e., the effective counts, stayed nearly constant for all the projection images. The rotated diagonal algorithm was applied to each projection image for pincushion distortion correction. To avoid the effect of reflected photons from the edges of the CsI:Tl scintillator, we further reduced the FOV of the detector to 6 mm $\times$ 6 mm. All events in each angular projection were binned into a 128×128 matrix (0.047 mm isotropic voxel size) before reconstruction.

VII. Digital Phantom and Expected Reconstruction Image

We created a digital phantom with a 128×128 matrix as shown in Fig. 6(a) in order to precisely compare the expected reconstruction result with the experimental results. The dark region illustrates the expected radioactivity distribution within a plane through the mouse heart phantom and the dotted rectangular region shows the wall defect. This image needs to be blurred with intrinsic spatial resolution limit of the pinhole and imaging system in order to be able to compare its image quality to the image reconstructed from experimental data. This blur was achieved by convolving a Gaussian function with the measured FWHM of the PSF of the PSAPD-based gamma camera at -25°C from Fig. 1. The FWHM was 0.74 mm. Fig 6(b) shows the blurred theoretical ideal image. A small rectangle on the left side in the Fig. 6(b) is the defined region for the calculation of a circumferential profile and illustrates the activity distribution away from the center of the annulus as a function the angle.

VIII. Reconstructed Image Comparison

We used a cone-beam iterative maximum likelihood expectation maximization (MLEM) algorithm to reconstruct the mouse heart phantom with 10 iterations [19]. Attenuation correction was not used, as the attenuation for 140 keV radiation through the thin 1 mm walls of the acrylic phantom is only $\sim 2\%$. Fig. 7(a) shows the short-axis image of the mouse heart phantom for projection data taken at -25°C . The center rod and the location of the defect are clearly visible on the reconstructed image. Fig. 7(b) illustrates the same reconstructed short-axis image for the projection data taken at 5°C . Fig. 7(c) shows the circumferential profile of the stacked (5 slices, 0.23 mm thick) short-axis image (Fig. 7(a), at -25°C), where the non-uniform activity distribution due to the defect is clearly observable at ~ 280 degrees. The circumferential profile of the expected blurred image shown in Fig. 6(b) is also plotted in Fig. 7(c) and 6(d), along with the circumferential profile of the stacked short-axis image in Fig. 7(a) and 7(b). Different FWHM widths of Gaussian blur were applied for -25°C (FWHM = 0.74 mm) and 5°C (FWHM = 0.88 mm intrinsic resolution) to the ideal digital mouse heart phantom. The image taken at a higher temperature (5°C) has increased fluctuation as shown in Fig. 7(d), caused by the increased temperature-dependent dark current shot noise and possible gain variation over the PSAPD surface, which changes with different temperature. In addition, we measured the SNR by defining three regions of interest (ROI) over the normal myocardium as shown in Fig. 6(a), noted as L, U, and R, which are located inside the area filled with radioactivity. The SNR was calculated as

$$SNR = \frac{\bar{\chi}}{\sigma}, \quad (1)$$

where $\bar{\chi}$ is the mean value of activity in the ROIs, and σ is the standard deviation of activity in the defined ROIs. The calculated values of SNR for ROIs at three different temperatures are as shown in Table I. For ROI U and R, the reconstructed image at -25°C shows the best SNR, and for ROI L, the SNR of the image measured at 5°C is slightly higher than the others. This result shows that there is small but quantifiable signal degradation for the image

taken at a higher (above 0°C) temperature. No post-filtering was applied to the reconstructed image, which may have contributed noise in addition to the statistical noise from the typical MLEM reconstruction algorithm.

IX. Discussion

Dark current shot noise can be suppressed by operating the PSAPD at lower temperatures [20]. Typically, operating the PSAPD at lower temperatures results in a slight increase in gain at a fixed bias voltage. However, at higher temperatures, a larger bias voltage (i.e., breakdown voltage) can be applied to the PSAPD resulting in a higher gain and potentially higher SNR [16],[21],[22]. In addition, the scintillation light yield for CsI:Tl varies with temperature [23]. From -25 °C to 5 °C, there is approximately 5% decrease in light yield. Thus, we need to find the operating temperature and bias voltage to maximize the PSAPD and CsI:Tl SNR.

While the image quality of the PSAPD-based camera degrades with increased temperature, it can be used for high resolution (submillimeter) small animal SPECT imaging applications at temperatures of 5 °C. We were able to achieve image quality that clearly shows the structure of the phantom, and the size and location of the defect, using our particular PSAPD at a temperature of 5 °C. Further enhancement of the reconstructed images might be achieved by digitally filtering out noise from the images or by using attenuation and scattering corrections in the iterative reconstruction, potentially allowing operation at temperatures higher than 5 °C.

A major limitation for commercialization of PSAPDs for use as photodetectors in SPECT scanners has been the low operating temperature and the associated cooling systems necessary for dark current shot noise. We have demonstrated that PSAPDs can exhibit good performance at relatively high temperatures, opening up the possibility of using other cooling methods for the PSAPD-based detection systems and avoiding problems with water condensation. Simpler cooling methods, such as thermoelectric coolers (TECs) can be used, eliminating the need for vacuum systems as water condensation would not be an issue. An efficient liquid chiller is also a good candidate, but high thermal conductance is still required between liquid chiller and PSAPD. A commercially available air-chiller might be an effective way of cooling, considering its simplicity and availability.

The temperature dependence of the PSAPD detector was examined by observing the visibility of the pixellated scintillator segmentation. Scintillation crystals with smaller segmentation sizes would in principle produce a more accurate measurement of the intrinsic PSAPD noise and spatial resolution; however, it is difficult to fabricate a scintillation crystal with smaller segmentation (< 400 μm) [24].

Multiple PSAPDs can be tiled together to form a large area detector [10]; however, the reflected photons at the edge of the scintillator and dead space between PSAPDs remain as problems. In order to solve these problems, the reflected photons (bias build-up at the edge) can be subtracted at the projection level, and the dead space can be compensated by statistic-based positioning algorithm [25],[26].

X. Conclusion

We experimentally characterized the temperature dependence of the image quality of a PSAPD-based gamma camera with intrinsic position sensing capability, submillimeter spatial resolution, and a compact size. The flood image obtained using a segmented, CsI:Tl crystal, PSF measurements, and uniformity variation measurement made with a continuous 2-mm thick CsI:Tl scintillator showed image quality illustrating clear distinction of phantom

structure and the defect at -25°C with -1640 V applied. Reconstructed SPECT images of a mouse heart phantom also showed slightly worse image quality at 5°C than at -25°C . However, the structure of the phantom and the size and location of the defect were clearly observable at 5°C . This warrants investigating the use of simpler cooling methods for a PSAPD-based SPECT imaging system.

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Fig. 1. Photographs of (a) the continuous CsI:Tl scintillator and (b) the fabricated PSAPD with an 8mm $\times$ 8mm active area and 4 anodes and 1 cathode.

Fig. 2.

FWHM of the PSF versus temperature with error bars. The FWHMs are in the range of 0.6 mm to 1.1 mm throughout the temperature variation from -40°C to 20°C , with the small FWHMs found in the region from -30°C to -20°C . The dotted line shows the fitting result indicating slight increase (slope=0.0018) in PSF FWHM versus temperature.

Fig. 3.

The mean-to-standard deviation ratio of counts inside the flood image is plotted versus temperature with error bars. The best uniformity is at -25°C as shown. The dotted line shows the fitting result indicating degradation of uniformity quality with increasing temperature.

Fig. 4. Flood images taken with the new PSAPD coupled with segmented CsI:Tl scintillator and $^{99\text{m}}\text{Tc}$ -pertechnetate source at various temperatures (a) -40°C , (b) -25°C and (c) 5°C

Fig. 5. Schematic drawing of the PSAPD-based gamma camera setup for the mouse heart phantom study. The inset shows a three-dimensional drawing of a mouse heart phantom with a defect.

Fig. 6.

(a) Short-axis view of the digital mouse heart phantom with defect. Three different ROIs drawn over the normal myocardium are shown. (b) Expected reconstructed image of the digital mouse heart phantom after applying Gaussian blur. Rectangular area with $1 \text{ mm} \times 0.7 \text{ mm}$ dimension was defined for circumferential profile calculation.

Fig. 7.

Reconstructed images and circumferential profiles from those images at two different temperatures. (a) Stacked short-axis reconstructed image of the mouse heart phantom for data taken at -25°C . The location of the defect is clearly visible, with less concentrated activity compared with the rest of activity distribution. (b) Stacked short-axis reconstructed image of the mouse heart phantom at 5°C . (c) Circumferential profile of the mouse heart phantom slice shown in (a), along with the profile of the expected image shown in Fig. 5(b). (d) Circumferential profile of the mouse heart phantom slice shown in (b), along with the profile of the expected image shown in Fig. 6(b).

TABLE I

Temperature dependent noise in reconstructed image

		-40°C	-25°C	5°C
L	$\bar{\chi}$	0.99	0.89	0.91
	σ	0.22	0.18	0.18
	$\bar{\chi} / \sigma$	4.50	4.94	5.06
U	$\bar{\chi}$	0.90	0.92	0.75
	σ	0.19	0.18	0.18
	$\bar{\chi} / \sigma$	4.74	5.11	4.17
R	$\bar{\chi}$	0.97	1.05	1.07
	σ	0.21	0.16	0.22
	$\bar{\chi} / \sigma$	4.62	6.56	4.86