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Count Rate Studies of a Box-Shaped PET Breast Imaging System Comprised of Position Sensitive Avalanche Photodiodes Utilizing Monte Carlo Simulation

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Abstract

We are investigating a high-sensitivity, high-resolution positron emission tomography (PET) system for clinical use in the detection, diagnosis and staging of breast cancer. Using conventional figures of merit, design parameters were evaluated for count rate performance, module dead time, and construction complexity. The detector system modeled comprises extremely thin position-sensitive avalanche photodiodes coupled to lutetium oxy-orthosilicate scintillation crystals. Previous investigations of detector geometries with Monte Carlo indicated that one of the largest impacts on sensitivity is local scintillation crystal density when considering systems having the same average scintillation crystal densities (same crystal packing fraction and system solid-angle coverage). Our results show the system has very good scatter and randoms rejection at clinical activity ranges (~200 μCi).

KEYWORDS: .

1. Introduction

To evaluate the performance of new detector systems used in applications such as high-energy physics and nuclear medicine, Monte Carlo is one of the first tools utilized [1-4]. Sensitivity, count rate performance, and reconstructed spatial resolution are a few of the characteristics of the modeled system that can be measured without having to commit the time and money to build every configuration under consideration. The modeling capabilities of a Monte Carlo package is verified by comparing simulation results with those obtained from built systems. In the case of GATE, these systems include ecat exact HR + (CPS), Allegro (Philips), GE Advance (GEMS), Micropet P4 (Concorde) and Micropet Focus (Concorde) for PET systems alone.

The purpose of this study was to determine the factors affecting count rate performance of a positron emission tomography system dedicated to imaging the breast using Monte Carlo techniques.

2. System Configuration

The simulated position sensitive avalanche photodiodes (PSAPDs) were modeled based on 11 mm × 11 mm × 230 μm prototypes from RMD, inc. An 8 × 3 array of 1mm × 1mm × 3mm LSO scintillation crystals were sandwiched between a reflector on one side and optically coupled to the PSAPDs on the sensitive 8mm x 8mm area on the other. The prototype PSAPDs are mounted on a thin (50 μm) flex cable for signal acquisition and delivering high voltage to the semiconductor. The total thickness of the flex cable, PSAPD, crystals and reflective film was 1.3 mm. A module, shown in Figure 1, is made of two PSAPDs (and associated crystal and reflector) attached to one flex cable. The crystal array-PSAPD structures were configured edge-on with respect to incoming 511 keV photons and were repeated trans-axially to form the four heads in the investigated geometry. This four-headed or 'box' system is further studied here because it was found to be the configuration with the highest sensitivity [5].

From previously reported prototype measurements with the LSO-PSAPD detectors [6-8], we have achieved excellent spatial (~1 mm fwhm), energy (~10-12% fwhm at 511 keV) and coincidence time (~2ns fwhm) resolutions. The software package,

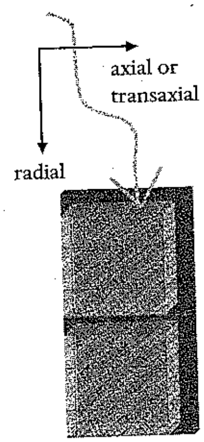

Fig. 1. Module design: a 1x2 array of PSAPDs (black) aligned edge-on to the incoming 511 keV photon, each PSAPD having an 8x3 array of LSO crystals (grey).

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GATE, a recently developed add-on to the Geant4 high-energy package, was used to model the physics of the processes involved in PET detection [4, 9]. In all simulations, energy resolution was set at 12% of 511keV, time resolution at 2ns, and a paralyzable dead time of 5us was applied to each module. Image reconstruction taking the depth of interaction into account has not yet been performed, these data will focus on factors affecting count rates and signal to noise estimations.

3. System Simulations

3. 1. *NEC and Count Rate*

Signal to noise (SNR) was estimated for the system using the NEMA (National Electrical Manufacturer's Association) standard: noise equivalent counts (NEC). The following is the conventional formula for NEC:

$$\mathrm{NEC} = \frac{T^2}{T + S + 2R} \qquad \text{(eq. 1)}$$

where T, S and R are true, scatter and random events respectively. NEC rates were calculated for the worst-case scenario of a uniformly filled cylindrical phantom, with diameter 16cm and length 12cm, which almost completely fills the detector inner volume (Figure 2). A report shows the average uncompressed breast thickness to be 7.4cm, the length to be 12.1cm, and the prone lateral width to be 15.4 cm [10] for patients undergoing scintimammography imaging. Since the volume of the phantom used in the NEC measurements is over two times the average imaged breast size, and about equal to the largest breast volume (in the FOV) in the study, we would expect the fraction of scatter in these measurements are higher than nearly all patients, giving a upper-bound on scatter rates. We also calculated an upper-bound on random rates by using the NEC formula which compensates for a delayed-coincidence measurement. The delayed-coincidence measurement is used to estimate randoms rates on-line in a detector system, but since we are using Monte Carlo, we in fact know these events precisely. This consideration increases the random contribution to nec two-fold. Therefore, we believe these are conservative values for all of the scatter, true and NEC rate measurements.

The amount of injected activity that accumulates in the breast is usually around 200 μCi. The other regions of high uptake such as the heart and lungs that contribute to background are not simulated in this study. The T, S, and R rates were obtained using a cylinder uniformly filled with water and 200 μCi of ^{18}F, the isotope used in the common PET tracer fdg (fluoro-deoxy-glucose).

Plotted in Figure 3 for each rate; NEC, R, S and T respectively; are: (left) a three dimensional plot of the rate surface by varying coincidence time window (CTW) and energy window (EW) and two line plots of the rates by varying (middle) EW for various CTWs and (right) CTW for various EWs. The maximum SNR, as measured by NEC, was observed for a 24% energy window, and a 63 or 8ns time window.

B. Count Rate vs. EW, CTW and Concentration

To assess the count rate performance of the system, we use the same 16cm diameter cylindrical water phantom described in the previous section and filled it uniformly with a range of activities of ^{18}F that include those pertinent to the application of breast imaging.

Plotted in figure 4 is the system's response from 10 μCi to 8 mCi of activity for three energy windows: 12%, 24%, an 36% of 511keV, each plot having a coincidence time window of 8ns. T, S, R, and NEC are shown for each activity as their fraction of total detected event rate, i.e., T+S+R = 1 for any activity. The total event rate was normalized to one to plot it using the same ordinate axis values. The peak total rate is given in the title of each of the graphs.

4. Discussion and Conclusions

To determine how best to construct a PET detector system constructed from LSO-PSAPD modules dedicated to breast imaging, we previously used GATE to choose between possible system configurations based on sensitivity measurements alone. In order to assess signal to noise ratio, as measured by NEC, we varied the coincidence time window (CTW) and energy window (EW). Since SNR translates into detectability of lesions in a reconstructed image, the peak NEC location on this CTW-EW surface should provide operation parameters for the data acquisition system. We found 24% energy window, twice the energy resolution, and 6ns or 8ns time window,

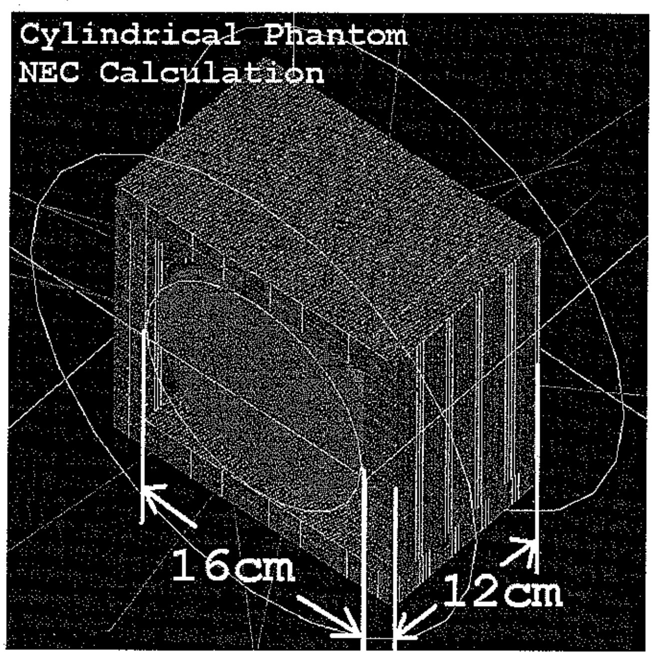

FIG. 2. An OpenGL rendering of the box system and cylindrical breast phantom used in GATE simulations.

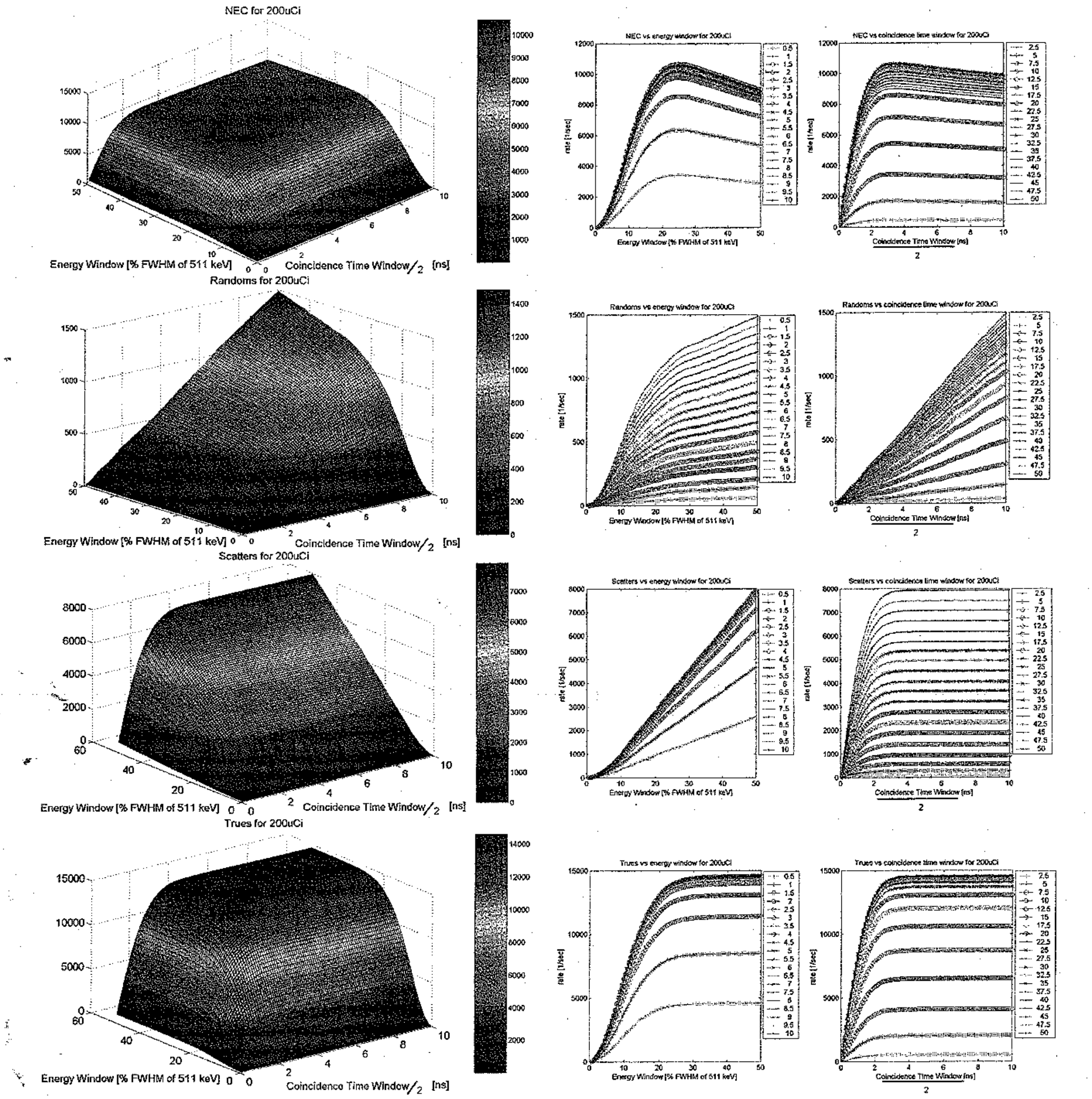

FIG. 3. 3d surfaces (left) and 2d plots (middle and right) of NEC, Random, Scatter, and True count rates vs. energy window and coincidence time window.

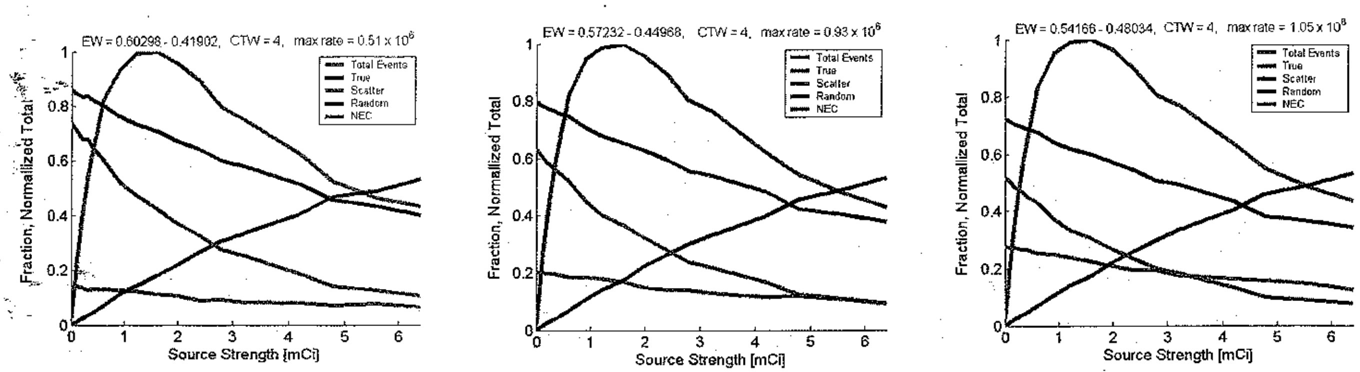

FIG. 4. Count rate vs. concentration for various energy windows: a) EW: 12% CTW: 4ns, b) EW: 24% CTW: 4ns, c) EW: 36% CTW: 4ns. The total counts are plotted normalized to the peak total counts. The trues, scatters and randoms are plotted as their fraction of the total. NEC is also scaled to the total counts.

3 or 4 times the coincidence time resolution, to give the best NEC rate. Looking at system performance in the activity range in the application of dedicated breast imaging (a few hundred micro-Curie), we see that the slope of total count rate vs. activity is still nearly linearly increasing, so we are not operating near the system counting limitations due to multiples rejection and module dead-time.

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