Abstract—Geiger-mode avalanche photodiode (GAPD) has been demonstrated to be a high performance PET sensor because of high gain, fast response, low excess noise, low bias voltage operation and magnetic field insensitivity. The purpose of this study is to develop a PET for human brain imaging using $4 \times 4$ array of large size GAPD. PET detector modules were designed and built to develop a prototype PET. The PET consisted of 72 detector modules arranged in a ring with an inner diameter of 330 mm. The LYSO arrays consisted of $4 \times 4$ array of $3 \times 3 \times 20$ mm$^3$ pixels, which were 1-to-1 coupled to $4 \times 4$ arrays of 9 mm$^2$ GAPD pixels (SensL, Ireland). The GAPDs were tiled together using flip chip technology on glass and operated at a bias voltage of 32 V for a gain of $3.5 \times 10^6$. The signals of the each module were amplified by a 16 channel preamplifier circuit with differential outputs and then sent to a position decoder circuit (PDC), which readout digital address and analog pulse of the one interacted channel from 64 signals of 4 preamplifier boards. The PDC output signals were fed into FPGA-embedded DAQ boards. The analog signal was sampled with 100 MHz, and arrival time and energy of the digitized signal were calculated and stored. The coincidence data were sorted and reconstructed by standard filtered back projection. The energy and time resolution of LYSO-GAPD block detector for 511-keV was 20.4% and 2.4 ns, respectively. The developed PDC could accurately provide the interacted PET signal and reduce the number of the readout channels of PET detector modules based on array type GAPD. The rods down to a diameter of 3.5 mm were resolved in hot-rod phantom image acquired with the brain PET which is similar to the image obtained by Monte Carlo simulation. Activity distribution pattern between white and gray matter in Hoffman brain phantom was well imaged. These results demonstrate that high performance PET could be developed using the GAPD-based PET detectors, analog and digital signal processing methods designed in this work. The prototype brain PET will be tested in a clinical 3T MRI to construct a combined PET-MRI.

I. INTRODUCTION

POSITRON Emission Tomography (PET) is a powerful molecular imaging modality that uses positron-emitting radionuclides to provide exceptionally sensitive assays of a wide range of biological processes [1-3]. Several studies have been reported on the fabrication of PET system based on avalanche photodiode (APD) instead of the photomultiplier tube (PMT) for MRI compatible PET [4-6]. However, APD has relatively large excess noise, low gain (in the range of 100–100,000) and poor timing capabilities [7, 8]. In recent years, Geiger-mode avalanche photodiode (GAPD) has attracted interests for its use as scintillator readout in PET applications [9]. GAPD consists of an array of avalanche photodiode micro-cells operated in Geiger-mode with the output summed from all the micro-cells, each with an integrated quenching resistor. It provides a number of advantages over APD detectors including; high gain ($\sim 10^6$), low bias voltage and fast timing response [10].

II. PURPOSE

- Develop a PET system consisting of 72 detector modules, 18 position decoder circuits (PDC) and 3 FPGA based DAQ boards for human brain imaging using $4 \times 4$ array of large size GAPD
  - Assess 80 LYSO crystal arrays to ensure the uniform performance of each crystal pixel
  - Evaluate the performance of GAPD arrays for application as PET photosensor
  - Evaluate the effect of gain correction by gain adjust circuit implemented in the PDC
  - Estimate the performance of the designed PET using Monte Carlo simulation method

- Evaluate the performance of the prototype brain PET including imaging capability of hot-rod and Hoffman brain phantoms

III. MATERIALS AND METHODS

A. PET detector module

- Detector module components (Fig. 1)
  - Scintillator:
    - Cerium-doped Lutetium Yttrium Orthosilicate (LYSO)
4 × 4 matrix of 3 × 3 × 20 mm³ crystals (Sinoceramics, China)
- 3.3 mm crystal pitch, polish treatment on all sides of crystal
- Optically isolated with a 0.3 mm white epoxy resin

- Photosensor:
  - Geiger-mode avalanche photodiode (GAPD) (SensL, Ireland)
  - 3-side scalable structure
  - 3.5 × 10⁶ gain at 32 V low bias voltage
  - 4 × 4 array of 2.85 × 2.85 mm² pixels, 3.3 mm pixel pitch
  - 3640 microcells per pixel, 35 μm microcell

Fig. 1. 4 × 4 matrix LYSO crystals and a 4 × 4 array GAPD used to construct PET detector module

- Components coupling and packing (Fig. 2)
  - Coupling: direct coupling LYSO to GAPD without optical-coupling material
  - Packing: each detector module independently encapsulated from light

Fig. 2. LYSO-GAPD detector assembly used in this study

B. Analog and digital signal processing

- Connection between PET detector module and preamplifier
  - Use flexible flat cable (FFC) of 300 cm long (Fig. 3)
  - Purpose: to separately locate PET detector module (inside MR bore) and preamplifier (outside MR bore)

Fig. 3. PET detector module and preamplifier connected using flexible flat cable of 300 cm long

- Preamplifier (SensL, Ireland)
  - Amplify 1,000 times the detector module signals
  - Supply voltage: 5.2 V
  - Output impedance: 47 Ω
  - Output signal characteristics (at 511 keV γ-ray)
    - Rise/Fall time: 30/170 ns
    - Amplitude of differential output signals: 250 mV

- Position decoder circuit (PDC) (Custom made)
  - Readout channel address and analog pulse of the channel interacted with coincidence event among 64 preamplified signals transmitted from 4 detector modules (Fig. 4)

Fig. 4. Configuration of position decoder circuit

- Data acquisition module (DAQ) [13]
  - Hardware: VHS-ADC Virtex-4 boards (Lyrtech, Canada)
  - Sampling by 105 MHz ADC
  - Calculate energy and arrival time of the digitized signal by initial rise interpolation method implemented using FPGA [14]
  - Store the data in list mode format

C. Performance Evaluation of PET Components

1) Assess LYSO array performance

- Experimental configuration
  - Purpose: to ensure the uniform performance of each crystal pixel of 80 4 × 4 LYSO arrays
  - 1 PET detector module + 1 preamplifier + 1 DAQ module (Fig. 5)
  - Radiation source: point source of 220 kBq Na-22
  - Duration of data acquisition: 5 min (per crystal block)

Fig. 5. Configuration of test set-up for LYSO array selection
Measurement
- Acquire energy spectra of 80 $4 \times 4$ LYSO arrays to compare photopeak position and energy resolution
- Compare crystals located at a same sensor pixel to rule out variability originated from sensor and to examine the variability only from LYSO crystals
- Calculate means of energy resolution and photo peak position using the energy spectrum of 80 crystal pixels of the same position among 16 pixels of an array

Selection range [11]
- The specification is as below list. If any crystal is out of the range, the crystal block was exchanged.

<table>
<thead>
<tr>
<th>Parameter @ 511 keV</th>
<th>Specification</th>
</tr>
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<tbody>
<tr>
<td>Photo peak position</td>
<td>Mean $\pm$ 23% of mean value</td>
</tr>
<tr>
<td>Energy resolution</td>
<td>Mean $\pm$ 31% of mean value</td>
</tr>
</tbody>
</table>

2) Evaluate the basic performance of GAPD array

Measurement [12]
- Acquire energy spectrum, count rate, flood histogram and variation of 511-keV photopeak channel using one PET detector module
- The resulting flood histogram was analyzed in terms of uniformity given by the ratio between minimum and maximum counts and variation of the 511 keV photopeak position for all 16 channels.

$$\text{Count uniformity} = \frac{\text{Count}_{\text{min}}}{\text{Count}_{\text{max}}}$$

$$\text{Variation}_{\text{photopeak}} = \frac{\text{standard deviation}}{\text{average of photopeak channels}} \times 100(\%)$$

- Acquire coincidence timing spectrum using a pair of PET detector modules

3) Evaluate the effect of gain correction in the PDC

Experimental configuration
- Gain adjustment in PDC: analog circuit capable of adjusting DC offset level and gain of the PET detector output signals (Fig. 6)
- Adjusting the value of a variable resistor connected in series to the output of offset level translation circuit.
- Gain range: 1 to 2

PET PET System

PET scanner configuration (Fig. 7)
- Detector module number: 72 unit
- Ring diameter: 330 mm
- Axial field-of-view: 12.9 mm

PET electronics configuration
- Preamplifier number: 72 unit
- PDC: 18 unit
- DAQ (VHS-ADC Virtex-4 board): 3 unit

Performance Estimation of the Brain PET Using Monte Carlo Simulation

Simulation parameters (Fig. 8)
- Simulation tool: GATE (Geant4 Application for Tomographic Emission)
- Point source:
  → Size: 1 mm diameter
  → Activity: 3.7 MBq
  → Source location: move from center to 100 mm off-center, in 20 mm steps
- Hot-rod phantom
  → Rod diameter: range from 2.5 to 6.5 mm
  → Activity: 13 MBq
- 3D digital Hoffman brain phantom
  → Activity: 96 MBq
- Energy window: 250 to 650 keV
- Image reconstruction: 2D FBP

- Measurement
  - Draw radial profiles of reconstructed images of point source as a function of source location to evaluate the spatial resolution over the FOV
  - Simulate hot-rod and 3D Hoffman brain phantom images to evaluate the imaging performance of the designed PET

**F. Performance Measurement of the Prototype PET**

- Experimental configuration (Fig. 9)
  - Detector: 72 detector modules consisted of 4x4 array of LYSO-GAPD
  - Cable length between detector module and preamplifier: 300 cm
  - Hot-rod phantom
    → Rod diameter: range between 2.5 and 6.5 mm
    → Radiation source: 74 MBq F-18
    → Total acquisition counts: 1 million
  - Hoffman brain phantom
    → Radiation source: 55 MBq F-18
    → Total acquisition counts: 6.8 million
  - Energy and time windows: 250 to 650 keV, 60 ns
  - Image reconstruction: 2D FBP
  - Image corrections: normalization, random

**IV. RESULTS**

A. Performance Measurement of LYSO Crystal

Average photopeak position and energy resolution of one pixel at the same position in the 80 crystal blocks were 426.7 and 21.7%, respectively. The values of the most crystals were within the selection range, however, some of pixels were out of selection range (Fig.10). These crystal blocks having abnormal pixel were exchanged to normal crystal.

![Fig. 10. Photopeak position (a) and energy resolution (b) of one pixel at the same position in the 80 crystal blocks](image-url)
B. Performance Measurement of the GAPD Array

TABLE II. PERFORMANCE MEASUREMENTS OF THE GAPD ARRAY

<table>
<thead>
<tr>
<th></th>
<th>Average energy resolution (n=16)</th>
<th>Average count rate</th>
<th>Flood histogram uniformity</th>
<th>Variation of 511 keV photopeak channel</th>
<th>Timing resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>20.4±3%</td>
<td>530 cps/µCi</td>
<td>1:1.5</td>
<td>7.6%</td>
<td>2.4 ns</td>
</tr>
</tbody>
</table>

Fig. 11. Flood histogram of data acquired with 4 x 4 array of LYSO-GAPD

C. Gain Correction

The ratio of the highest and lowest photopeak positions was improved from 1.8 to 1.0 after gain correction by using the gain adjust circuit implemented in PDC (Fig. 12, 13).

Fig. 12. Relative photopeak position before and after the gain correction

Fig. 13. 32 channel energy spectra (Na-22) obtained after gain correction by using the gain adjust circuit

D. Performance Estimation of the Brain PET Using Monte Carlo Simulation

Radial resolution of reconstructed point source image was degraded from 3.3 mm to 6.1 mm, at a 100 mm off-center (Fig. 14).

Fig. 14. Reconstructed images of the point source obtained by moving a source from the center to the radial offset of the scanner’s FOV, at 20 mm intervals (left), and their radial resolutions (right)

E. Performance Measurement of the Prototype PET

PET images representing activity distribution patterns of hot-rod and Hoffman brain phantoms were successfully acquired using the brain PET developed in this study (Fig. 15). The rods down to a diameter of 3.5 mm were resolved in hot-rod phantom image acquired with the brain PET which is similar to the image obtained by Monte Carlo simulation. Activity distribution pattern between white and gray matter in Hoffman brain phantom was well imaged.

Fig. 15. Hot-rod (top row) and Hoffman brain phantom (bottom row) images acquired with simulation and with the brain PET developed in this study

V. SUMMARY AND CONCLUSION

The results of this study demonstrate that the GAPD array can provide good energy resolution, count rate, timing resolution and flood histogram uniformity as a PET detector. Gain uniformity for all channels of the PET detector modules was improved 80% by using the gain correction circuit.
implemented in PDC. The developed PDC could accurately provide the interacted PET signal and reduce the number of the readout channels of PET detector modules based on array type GAPD. PET images were successfully acquired using the prototype brain PET consisting of 72 detector modules and these results demonstrate that high performance PET could be developed using the GAPD-based PET detectors, analog and digital signal processing methods designed in this work. The prototype brain PET will be tested in a clinical 3T MRI to construct a combined PET-MRI [15, 16].

REFERENCES