



Contents lists available at ScienceDirect

Nuclear Instruments and Methods in Physics Research A

journal homepage: www.elsevier.com/locate/nima

Development of large-area, reverse-type APD-arrays for high-resolution medical imaging

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ARTICLE INFO

Available online 31 January 2009

Keywords:

Avalanche photodiode

γ -rays

Scintillation detection

ABSTRACT

Avalanche photodiodes (APD) offer advantages in terms of weak scintillation detection, fast time response, and magnetic field insensitivity. We have developed new types of large-area, reverse-type APD-arrays specifically designed for high-resolution positron emission tomography (PET). Each device has a monolithic 16×16 (or 8×8) pixel structure with an active area of 1.0 (or 4.0, 0.25) mm^2 for each pixel. An excellent gain uniformity ($\leq 10\%$) and low dark noise (≤ 0.3 nA) have been achieved, measured at room temperature. Energy resolution of 7.2% (FWHM) was obtained for the direct detection of 5.9 keV X-rays, while 10.2% (FWHM) was obtained for 662 keV gamma-rays when coupled with a LYSO scintillator matrix. An excellent time resolution of 102 ps (FWHM) was obtained for a monolithic, $3\text{ mm}\phi$ APD pixel. These results suggest that APD-arrays could be a promising device for future applications in nuclear medicine.

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1. Introduction

The avalanche photodiode (APD) is a compact, high-performance light-sensor recently applied in various fields of experimental physics. In particular, the reverse-type APD offers great advantages in detecting weak light scintillation signals, thanks to its narrow high-field multiplying region close to the front end [1–3]. Moreover, this type of APD works at relatively low bias voltage (300–400 V) and achieves excellent noise characteristics. Unfortunately, only a single device (S8550; a monolithic 8×4 pixel structure with an active area of 2.56 mm^2 for each pixel) is now commercially available for application to high-resolution scintillator matrices read-out [4]. This paper reports on the development of new types of large-area reverse-type APD-arrays, specifically designed for future use in a APD-based PET (positron emission tomography) scanner.

PET scanners are powerful tools used for the study of physiological processes in vivo. Current developments aim at building smaller, and less expensive devices with improved image resolution (matching the sub-mm level or better [5]). The recent onset of using dual modality PET/CT imaging has had a profound

effect on clinical diagnosis in radiology, oncology and other areas of nuclear medicine. However, CT exhibits a poor soft-tissue contrast, and also subjects the patient to a significant radiation dose exceeding that received from PET itself. In contrast, magnetic resonance imaging (MRI) provides an excellent soft-tissue contrast and anatomical detail, without an additional radiation dose. In this sense, APD is of great interest because it is insensitive to the high magnetic field used in the MRI ($\sim 5\text{ T}$ [6,7]). Another important challenge for future PET detectors will be met by using time-of-flight (TOF) information to reduce statistical noise variance in the reconstructed image [8].

With these motivations in mind, a simple PET device consisting of an APD-array optically coupled with a LYSO scintillator is being fabricated. The signals from each pixel are read-out by a low-noise analog front-end ASIC specifically designed for the device [9]. The initial performance of the newly developed APD-arrays will be presented below in direct detection of soft X-ray photons, as well as in the read-out LYSO matrix.

2. Newly developed APD-arrays

Three types of large-area, reverse-type APD-arrays were made, that were developed based on the technology of the S8664 APD series (Hamamatsu). The basic characteristics of Hamamatsu

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Table 1
Parameters of newly developed Hamamatsu APD-arrays.

	TYP1	TYP2	TYP3
Matrix array	8 × 8	16 × 16	16 × 16
Pixel size (mm ²)	2 × 2	1 × 1	0.5 × 0.5
Pixel gap (mm)	0.3	0.4	0.4
Dark current ($M = 50$) I_D (nA)	0.5–1.3	0.1–0.3	0.1–0.4
Break-down voltage: V_{brk} (V)	379	376	380
Operation bias: $V_{M=50}$ (V)	355	333	356
Capacitance: C_{det} (pF)	13–15	4–5	3.3–4.6

All parameters are measured at +25 °C.

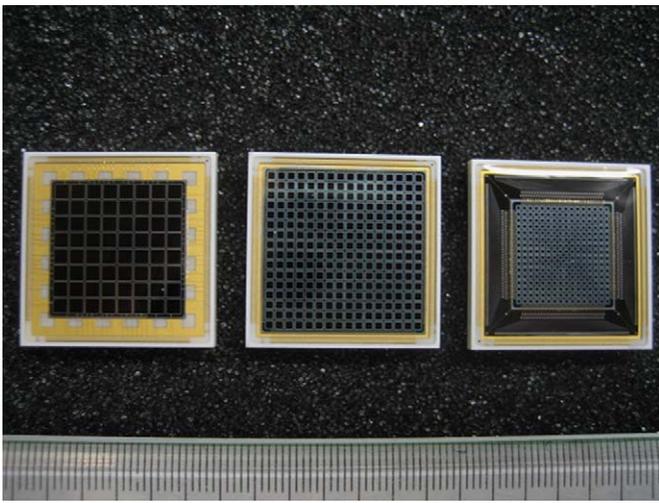


Fig. 1. A picture of the reverse APD-arrays developed in this paper. (TYP1 to 3; from left to right).

APDs are documented in detail in other literature [1–3]. Table 1 lists the design parameters, dark noise and gain characteristic of each device. TYP1 carries an 8 × 8 array of 2 × 2 mm² pixels, while TYP2 and TYP3 consist of a 16 × 16 matrix of 1 × 1 mm² (TYP2) and 0.5 × 0.5 mm² (TYP3) pixels, respectively. As shown in Fig. 1, all APD-arrays are embedded in a ceramic package of the same configuration, 27.4 × 27.4 mm² (1.27 mm pitch, pin-grid type).¹ We tried to maximize the effective area of the APD for a given geometry. For example, the TYP1 array has 0.3 mm gaps between the 2 × 2 mm² pixels, compared to 0.7 mm gaps for the 1.6 × 1.6 mm² pixels of S8550, thereby improving the effective area of the APD from 48% (S8550) to 76% (TYP1).

Our APD-arrays allow stable operations at gain up to 100, with extremely low dark noise for each pixel, even at room temperature. An avalanche gain of 50 is achieved with a bias voltage of 330–360 V, which is sufficiently low compared to that required for other types of APDs (as described elsewhere [10]). Furthermore the APD gain shows excellent uniformity over the entire APD pixels. Fig. 2 shows the gain variation (*left*) and dark noise distribution (*right*) for the TYP2 array, measured at gain $M = 50$ and $T = 25$ °C. Note the excellent uniformity of avalanche gain. Gain fluctuation is only at the 10% level over the APD device. Dark noise is also uniform with an average of $I_D = 0.23 \pm 0.08$ nA/pixel (1 mm²).

¹ At this stage, the ceramic package was specifically designed to be matching with TYP2 array (Fig. 1: center). We plan to make optimum packages also for TYP1 and TYP3 arrays in future fabrication to further reduce the dead space.

3. Detector performance

3.1. Direct X-ray detection

To demonstrate the performance of our APD-arrays, the device was irradiated by a ⁵⁵Fe source which emits 5.9 keV X-rays. Energy resolution of $7.2 \pm 0.6\%$ (FWHM) was obtained for all the APD pixels, representing one of the best records ever reported using APD devices (Fig. 3). Thanks to the extremely low dark noise, the energy threshold is as low as $E_{th} \sim 0.6$ keV, measured at room temperature (+25 °C). One drawback is that, the device does not allow efficient X-ray detection due to its thin depletion layer (~ 10 μm). For this particular purpose, thicker (≈ 130 μm) reach-through APDs and beveled-edge APDs are more advantageous as discussed in other literature [10,11].

3.2. Readout of LYSO scintillator matrix

Next a prototype gamma-ray camera consisting of an APD-array optically coupled with a LYSO matrix were fabricated. Fig. 4 (*upper*) shows a picture of the 16 × 16 LYSO matrix for TYP2, where each pixel is 1.3 × 1.3 × 10 mm³ in size and divided with the lattice of a thin reflective layer (3M ESR). Fig. 4 (*lower*) shows the TYP1 APD-array coupled with the LYSO matrix. In testing the proto-type, we used a low-noise analog front-end ASIC specifically designed for our APD–PET system [9]. Fig. 5 shows an example of an energy spectrum obtained with the TYP1 + LYSO array (measured at +25 °C) for the ¹³⁷Cs source. The energy resolution of the 662 keV gamma-ray is $10.2 \pm 0.2\%$ (FWHM). The variation in signal amplitude (due to inhomogeneities of APD gain and LYSO light yield) was only $\pm 16\%$ among 8 × 8 pixels. A small peak in the spectrum (arrow) is due to the cross-talk of scintillation light from neighboring LYSO pixels through the epoxy window covering the APD surface. This low-level interference is actually acceptable for PET imaging, but a revised version of APD-arrays with a thin epoxy coating is being fabricated.

3.3. Time response

Following a method described in other literature [12], a timing experiment on the reverse-type APD was carried out at the beamline NW2A of the PF–AR ring (KEK) at Tsukuba, Japan. In this experiment, we used a monolithic reverse-type APD pixel (S8664-30; Hamamatsu, 3 mmϕ) to simplify the setup. Note that the internal structure of the S8664-30 is exactly the same as that of the APD-arrays, except for the pixel geometry. A double-crystal silicon (111) monochromator was used to define X-ray energy as 16 keV with an energy resolution of $\Delta E \approx 4$ eV. The X-ray beam was focused by a mirror into a spot size of 0.7 × 0.3 mm². Output signals from the APD were processed with a constant fraction discriminator (CFD: ORTEC 935) and fed into a time-to-amplitude converter (TAC: TENELEC TC863) with a maximum range of 50 ns, then finally digitized by an analog-to-digital converter (ADC) with 2048 ch resolution (25 ps/ch). During the experiment, the beam at PF–AR was operated in single-bunch mode.

Fig. 6 shows the time response of the APD pixel as measured by the direct detection of 16 keV X-rays. The APD was operated under a bias voltage of 400 V, corresponding to an avalanche gain of ≈ 100 . A measured time resolution from raw data provides $\Delta T_{all} = 190$ ps (FWHM), while most of the broadening is due to a width of electron bunches of the light source itself, and estimated as $\Delta T_B = 160$ ps (FWHM). Hence the intrinsic time resolution of APD is expected to be $\Delta T_{APD} = 102$ ps (FWHM). Note that this is much better than the timing response reported in literature. For

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