Zero-dark-counting X-ray photon detection using a YAP(Ce)-MPPC detector and its application to computed tomography using gadolinium contrast media

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ABSTRACT

To measure X-ray spectra and to perform photon-counting computed tomography (PC-CT) with high count rates, we developed a zero-dark-counting spectrometer using a short-decay-time scintillator. A method exploiting a YAP(Ce) [cerium-doped yttrium aluminum perovskite] single crystal scintillator with a decay time of 30 ns and an MPPC (multipixel photon counter) has been developed to count X-ray photons. The photocurrent from the MPPC was amplified by a high-speed current-voltage amplifier, and the event pulse is sent to a multichannel analyzer (MCA) to measure X-ray spectra. The MPPC was driven under pre-Geiger mode at a bias voltage of the MPPC of 70.7 V and a temperature of 23°C. The PC-CT was accomplished by repeated linear scans and rotations of an object, and projection curves of the object were obtained by the linear

scan at a tube current of 1.0 mA. The exposure time for obtaining a tomogram was 10 min at a scan step of 0.5 mm and a rotation step of 1.0°. At a tube voltage of 100 kV, the maximum count rate was 200 kcps. In the PC-CT using gadolinium media, we observed image-contrast variations with changes in lower-level discrimination voltage of the event pulse using a comparator.

Keywords:

YAP(Ce)-MPPC detector

High-speed X-ray photon counting

High-speed amplifier

X-ray CT

High-speed comparator

Energy-dispersive effect

1. Introduction

To measure X-ray spectra, we usually use a cadmium-telluride (CdTe) detector because liquid nitrogen is not necessary to cool the CdTe detector. Therefore, we have developed energy-dispersive computed tomography (ED-CT) systems (Abudurexiti et al., 2010; Matsukiyo et al., 2011; Sato et al., 2012a) using CdTe detectors with an energy resolution of 1.2 keV (1%) at 122 keV to perform iodine (I) and gadolinium (Gd) K-edge imaging. Furthermore, a preclinical ED-CT system (Feuerlein et al., 2008; Schlomka, et al., 2008) has been developed to perform Gd-K-edge imaging using phantoms including Gd atoms. Successively, a silicon-PIN (Si-PIN) diode with an energy resolution of 0.15 keV (2.5%) at 5.9 keV is usually used to carry out X-ray fluorescence (XRF) analysis, and a high-energy-resolution ED-CT system (Hagiwara et al., 2011) has been developed and applied to 2 keV-bandwidth I-K-edge imaging. In addition, an XRF-CT system (Huo et al., 2008; Takeda et al., 2009) for measuring iodine density in mice has been developed using a germanium (Ge) detector and monochromatic sheet beams. Generally, with increases in the count rate of CT, the X-ray exposure time for CT can be reduced, and the image quality of tomography improves. However, it is not easy to increase the count rate to 1 Mcps (mega counts per

second) or beyond in the ED-CT system using conventional CdTe, Si-PIN and Ge detectors.

Recently, Mcps range photon-counting X-ray CT (PC-CT) systems have been developed using a multipixel photon counter (MPPC) module and short-decay-time scintillators. The decay times of an LSO [$\text{Lu}_2(\text{SiO}_4)\text{O}$] (Sato et al., 2011), a YAP(Ce) [cerium-doped yttrium aluminum perovskite] (Oda et al., 2011), and a zinc oxide [ZnO] (Sato et al., 2012b) single crystals are approximately 40, 30 and 1 ns, respectively, and the photon count rate has been increased to 15 Mcps using the ZnO crystal (Simpson et al., 2003; Sugimura, et al., 2011). However, the maximum dark count rate was 1 Mcps, and it was difficult to measure X-ray and γ -ray spectra.

To reduce the dark count rate of the MPPC, the bias voltage of the MPPC should be decreased to a value in pre-Geiger mode, and it is necessary to measure the boundary voltage between the pre-Geiger and Geiger modes. Because there are no readily available high-speed current-voltage amplifiers, a high-speed amplifier for the MPPC should be developed.

In our research, major objectives are as follows: development of a shielded YAP(Ce)-MPPC detector; development of a high-speed current-voltage amplifier; measurement of X-ray spectra at tube voltages below 100 kV using the detector; and performance of PC-CT using Gd media and a high-speed comparator. Therefore, we developed a high-speed photon-counting system to measure X-ray spectra using a YAP(Ce)-MPPC detector and performed PC-CT using Gd media.

2. Experimental setup

2. 1. Photon-counting using a YAP(Ce)-MPPC detector

Fig. 1 shows the block diagram for counting X-ray photons using a YAP(Ce)-MPPC detector. Although there are three types of 100, 400 and 1,600 pixels, we used a 400-pixel MPPC (S10362-11-050C, Hamamatsu) to measure the average photoelectric characteristics of the YAP(Ce)-MPPC detector. An X-ray generator (RXG-0152, R-tec) has been developed to produce low-dose-rate X-ray beams by decreasing tube current in the range from 1.0 μA to 2.0 mA.

The circuit diagram of a high-speed current-voltage amplifier is shown in Fig. 2. The

MPPC is driven under pre-Geiger mode at a bias voltage of 70.7 V and a temperature of 23°C. The X-ray photons from an X-ray tube are detected by the YAP(Ce)-MPPC detector, and the photocurrents flowing through the MPPC are amplified by the amplifier with an operational amplifier chip of AD8032. The event pulses from the amplifier are sent to a multichannel analyzer (MCA; MCA4000, γ PGT) to measure X-ray spectra with changes in the tube voltage and the insertion of a 3.0 mm-thick aluminum (Al) filter. In the detector, a YAP(Ce) crystal of $1.0 \times 1.0 \times 1.0 \text{ mm}^3$ is positioned on the light-receiving surface of the MPPC using epoxy resin and is covered with an Al cap with a 0.2 mm-thick Al window. The detector is shielded using both an Al case with a 25 μm -thick Al window and a BNC connector. The distance between the target in the X-ray tube and the detector is 1.00 m.

In order to measure γ -ray spectra with an energy resolution of 1% at 122 keV, we used a CdTe detector (XR-100T, Amptek). Fig. 3 shows relation between the γ -ray spectra measured by the CdTe and the YAP(Ce)-MPPC detectors using an americium-241 (^{241}Am) source. Using the YAP(Ce)-MPPC detector, although the peak energy of 59.5 keV could be observed, the energy resolution was determined to be approximately 100% at 59.5 keV.

2.2. PC-CT System

A PC-CT system is used for observing ED effect with changes in the lower-level discrimination voltage V_d of a comparator. A block diagram of the PC-CT system utilizing the detector is shown in Fig. 4. The CT system consists of the X-ray generator, a linear X-ray scanner based on the YAP(Ce)-MPPC detector with a scanning velocity of 25.0 mm/s, a turntable (SGSP-60YAW-OB, Siguma Koki), a two-stage controller (SHOT-602, Siguma Koki), a comparator device, a counter card (CC; CNT32-4MT, Contec), and a personal computer (PC). The distance between the X-ray source and the detector is 1.00 m and the distance from the center of the turntable to the scintillator is 40 mm to decrease the magnification ratio of the object. The oscillation-type linear scanner is composed of a scan stage (SGSP-26-100, Siguma Koki), the shielded YAP(Ce)-MPPC (YAP-MPPC13, Toreck) detector, and the amplifier. The scanning is conducted in both directions of its movement. The event pulses from the amplifier are

sent to the high-speed comparator for determining V_d , and the logical pulses are counted using the CC.

The circuit diagram of a comparator device is shown in Fig. 5. A comparator IC (integrated circuit) of AD8561 was used for determining V_d , which was controlled by a variable resistor of 10 k Ω and was measured using a direct-current (DC) voltage meter. We used an inverse-comparison circuit with a hysteresis line consisting of two resistors for preventing error oscillations from the IC output. Using this circuit, the initial comparator voltage before the input of the event pulse is approximately 4 V, and the voltage decreased from 4 to 0 V when the event pulse voltage increases beyond V_d . A condenser C_o of 1 nF is used to absorb voltage oscillations from a 50 Ω coaxial cable between the CC and comparator.

3. Results

3.1. X-ray dose rate and spectra

The measurement of X-ray dose is very important for inferring the skin dose for objects. The dose rate from the X-ray generator was measured using an ionization chamber (RAMTEC 1000 plus, Toyo Medic) placed 1.0 m from the X-ray source (Fig. 6). At a constant tube current of 1.0 mA, the dose rate increased with increasing tube voltage. The rates at tube voltages of 50 and 100 kV were 30.6 and 111 $\mu\text{Gy/s}$, respectively.

To measure X-ray spectra from the X-ray source with an energy resolution of 1% at 122 keV, we also used the CdTe detector. [The measured spectra using the CdTe detector are shown in Fig. 7.](#) Contrast media may be used to enable K-edge imaging and the Gd K-edge energy (50.3 keV) is also shown in the same figure for reference. X-ray photons with energies just beyond the K-edge are absorbed effectively by Gd atoms. With increase in the tube voltage, both the peak and the maximum energies increased. The peak energy increased by the insertion of the 3.0 mm-thick Al filter for absorbing low-energy photons.

The X-ray spectra measured by the YAP(Ce)-MPPC detector are shown in Fig. 8. The maximum energy did not correspond to the tube voltage owing to low energy resolution. The peak energy increased with increasing tube voltage and by the insertion of the Al

1
2 filter. At a tube voltage of 50 kV, the peak energy corresponded well to the
3 bremsstrahlung peak energy obtained using the CdTe detector. By the production of K-
4 series characteristic X-rays at a tube voltage of 100 kV, the peaks were approximately 5
5 kV higher than bremsstrahlung peaks obtained the CdTe detector. The maximum energy
6 increased with increasing tube voltage, and the maximum energy seldom varied with the
7 insertion of the Al filter.
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10 11 12 13 14 15 3.2. Event pulse and comparator output

16 Fig. 9 shows time relation between the event-pulse voltage and the comparator output
17 at a V_d of 0.8 V. The pulse height and the width were 2.2 V and 200 ns, respectively. The
18 initial voltage of the comparator output before the input of event pulse was 3.7 V, and
19 the comparator output $V_c(t)$ substantially decreased to 0 V when event-pulse voltage
20 $V_e(t)$ increased beyond V_d with a delay time of 10 ns; the pulse was then counted using
21 the CC. Successively, $V_c(t)$ increased from 0 to 3.7 V when $V_e(t)$ decreased below 0.8 V.
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30 3.3. Tomography

31 Tomography was performed at a scan step of 0.5 mm and a rotation step of 1.0° , and
32 the reconstructed relative maximum and minimum counts are automatically denoted as
33 black and white, respectively. On the other hand, tomograms are obtained as JPEG files,
34 and the maximum and minimum densities are defined as white and black, respectively.
35 The background density around the object was not normalized to the minimum density.
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41 Fig. 10 shows tomograms of two 15 mm-diam glass vials filled with two different
42 density Gd media (meglumine gadopentetate) of 15 and 30mg/ml, indicated as (a) and
43 (b) in the figure. At a tube voltage of 100 kV, the image density difference between two
44 media slightly decreased with increasing V_d from 0.1 to 1.0 V. On the other hand, the
45 image density difference increased with increase in the tube voltage from 50 to 100 kV
46 at a V_d of 0.1 V.
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52 Fig. 11 shows the density analysis of the tomograms in Fig. 10 using a program
53 Image J. At a tube voltage of 50 kV and a V_d of 0.1 V, the glass densities were quite high
54 compared with those of contrast media. When the tube voltage was increased from 50 to
55 100 kV at a V_d of 0.1 V, the image density difference between two media increased.
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However, the artifact between two vials was produced by the beam hardening at a tube voltage of 100 kV. The artifact improved with increasing V_d because the energy bandwidth decreased at a tube voltage of 100 kV.

The result of the comparison between the tomography of a rabbit-head phantom made using a YAP(Ce)-MPPC scanner and a radiography image is shown in Fig. 12. The blood vessels were filled with gadolinium oxide (Gd_2O_3) microparticles. The animal operation was carried out in accordance with the animal experiment guidelines of our university. Radiography (angiography) was performed with a flat-panel detector (FPD; Rad-icon Imaging 1024EV) to observe blood vessels. In radiography, fine blood vessels were observed because the pixel sizes are $48 \times 48 \mu\text{m}^2$. In tomography, the image density of Gd media seldom varied with changes in the V_d and the tube voltage, since the Gd microparticles in the vessels had high opacity in all the tomograms. On the other hand, the image densities of bones and muscles increased with decreasing tube voltage from 100 to 50 kV at a V_d of 0.1 V and with increasing V_d from 0.1 to 1.0 V at a tube voltage of 100 kV. Compared with radiography, the image contrast of vessels was high.

4. Discussion and conclusions

We performed both the measurement of spectra and the PC-CT using a YAP(Ce)-MPPC detector. The dark count rate was 0 cps with a bias voltage of the MPPC of 70.7 V. In the γ -ray counting using a ^{241}Am source and an MCA, the energy resolution of the detector was approximately 100% at 59.5 keV. The peak energy increased with increasing tube voltage and by the insertion of the Al filter. The pulse widths of the event pulse were approximately 200 ns, the width should be increased to beyond $5 \mu\text{s}$ to detect the event-pulse height correctly. Using an integrator with a time constant of $5 \mu\text{s}$ and an amplifier, long-width event pulses were obtained, and the spectra could be obtained. However, the characteristics concerning the energy resolution seldom varied, and it was difficult to increase the count rate of the event pulse.

Using the YAP(Ce)-MPPC detector, because it was difficult to amplify the photocurrent from the MPPC using readily available amplifiers, we developed a simple high-speed amplifier with a 80 MHz AD8032 chip.

In the PC-CT system, the image contrast of Gd media slightly fell with increase in the

V_d because the average photon energy increases with increasing V_d . Although it is difficult to perform the K-edge imaging using Gd and I media, it is possible to perform dual energy subtraction (DES) using a dual comparator device. Regardless of the energy resolution, the DES can easily be carried out by selecting two different-energy spectra using comparators. To perform the DES, it is important to obtain two different energy tomograms simultaneously to improve the image granulation caused by the statistical noises. In addition, a frequency-voltage converter with an integrator is very useful for improving the image granulation, and the image granulation improves with increasing time constant in the integrator.

The spatial resolution is primarily determined by the dimensions of the X-ray-receiving area of the YAP(Ce) scintillator, and the original dimensions were $1.0 \times 1.0 \text{ mm}^2$. To improve the spatial resolution, the dimensions should be reduced.

In this PC-CT system with a tube voltage of 100 kV, a current of 1.0 mA, and a V_d of 0.1 V, the maximum count rate was 200 kcps, and the photon count per measuring point was calculated as 4,000 counts (=200 kc/50). Furthermore, the rate substantially increased to beyond 500 kcps with a decrease in V_d and increases in both tube voltage and current.

The advantages of the PC-CT system are as follows: the noise shielding of the amplifier is not necessary; the photon count rate can be increased to 1 Mcps; and the exposure time for CT can be reduced by increasing scan velocity and by forming the detector array. On the other hand, a major disadvantage is a low energy resolution of 100% at 59.5 keV. In this experiment, because we used a 400-pixel MPPC, the energy resolution would be improved by increasing the pixel number to 1,600 pixels. Using a small-sized photomultiplier tube (H10721P-110, Hamamatsu) with a $2.5 \times 2.5 \times 1 \text{ mm}^3$ YAP(Ce) scintillator, because the resolution was 15% at 59.5 keV, the resolution of the YAP(Ce)-MPPC detector would also be improved by the development of a new I-V amplifier or by improving the photoelectric characteristics of the MPPC.

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Figure captions

Fig. 1. Block diagram for counting X-ray photons using a YAP(Ce)-MPPC. The photocurrent from the MPPC is amplified by a high-speed current voltage amplifier. The structure of the YAP(Ce)-MPPC detector is shown in the same figure.

Fig. 2. Circuit diagram of a high-speed current-voltage amplifier.

Fig. 3. Comparison between the two γ -ray spectra using a ^{241}Am source. (a) γ -Ray spectra using the CdTe detector and (b) γ -ray spectra using the YAP(Ce)-MPPC detector.

Fig. 4. Block diagram of a PC-CT system using a linear YAP(Ce)-MPPC scanner with a scan velocity of 25 mm/s. PC-CT is carried out by repeated linear scan and rotation of an object.

Fig. 5. Circuit diagram of a comparator device with a hysteresis circuit consisting of two resistors for preventing error oscillations from the IC output.

Fig. 6. X-ray dose rate measured using an ionization chamber placed 1.0 m from the X-ray source at a tube current of 1.0 mA.

Fig. 7. X-ray spectra measured using a CdTe detector at the indicated conditions. Gd K-edge energy of 50.3 keV is shown in the same figure for reference, and X-ray photons with energies just beyond 50.3 keV are absorbed effectively by Gd-based contrast media.

Fig. 8. X-ray spectra measured using the YAP(Ce)-MPPC detector at the indicated conditions. The maximum channel number increased with increase in the tube voltage. However, the maximum number was unaffected by the aluminum filtration.

Fig. 9. Time relation between event-pulse voltage and comparator output at a V_d of 0.8 V.

Fig. 10. Tomograms of two 15 mm-diam glass vials filled with two different density Gd media. The image density difference between two media increased with increasing tube voltage from 50 to 100 kV.

Fig. 11. Image density analysis of tomograms in Fig. 10 using a program Image J. The artifact improved with increasing V_d at a tube voltage of 100 kV.

Fig. 12. Tomography of a rabbit-head phantom. The blood vessels were filled with Gd_2O_3 microparticles. Image density of bones and muscles substantially increased with decreasing tube voltage from 100 to 50 kV at a V_d of 0.1 V.

Figure

Fig. 1

Fig. 2

Figure

Fig. 3

Figure

Fig. 4

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Fig. 6

Fig. 7

Fig. 8

Fig. 9

Scan step=0.5mm
Rotation step=1.0°
 V_d : Discrimination voltage
 V : Tube voltage

(a) 30 mg/ml Gd
(b) 15 mg/ml Gd

60 mm

$V_t=0.1$ V
 $V=50$ kV

$V_t=0.1$ V
 $V=100$ kV

$V_t=1.0$ V
 $V=100$ kV

15 mm-diam glass vials

Fig. 10

Figure

Fig. 11

Fig. 12