Dark-count-less photon-counting X-ray computed tomography system 
using a YAP-MPPC detector

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ABSTRACT
A high-sensitive X-ray computed tomography (CT) system is useful for decreasing absorbed dose for patients, and a 
dark-count-less photon-counting CT system was developed. X-ray photons are detected using a YAP(Ce) [cerium-doped 
yttrium aluminum perovskite] single crystal scintillator and an MPPC (multipixel photon counter). Photocurrents are 
amplified by a high-speed current-voltage amplifier, and smooth event pulses from an integrator are sent to a high-speed 
comparator. Then, logical pulses are produced from the comparator and are counted by a counter card. Tomography is 
accomplished by repeated linear scans and rotations of an object, and projection curves of the object are obtained by the 
linear scan. The image contrast of gadolinium medium slightly fell with increase in lower-level voltage (Vl) of the 
comparator. The dark count rate was 0 cps, and the count rate for the CT was approximately 250 kcps.

Keywords: YAP-MPPC detector, Sub-Mcps photon counting, dark-count-less detection, high-speed current-voltage 
amplifier, high-speed comparator, energy-dispersive effect

1. INTRODUCTION
Energy-selective X-ray imaging can be performed utilizing energy dispersion of X-ray photons, and several energy-
dispersive computed tomography (ED-CT) systems have been developed to carry out preclinical researches. 
Therefore, we have developed transmission-type ED X-ray cameras, reflection-type X-ray fluorescence (XRF) 
cameras, ED-CT systems, and XRF-CT systems. In particular, ED-CT systems have been applied to enhanced 
K-edge CT using iodine and gadolinium contrast media.

To reduce X-ray exposure time for ED-CT, we have developed photon-counting X-ray CT (PC-CT) systems using a 
multipixel photon counter (MPPC) module and short-decay-time single scintillator crystals. Using a YAP(Ce) [cerium-
doped yttrium aluminum perovskite] scintillator crystal, although the maximum count rate of the PC-CT system was 3 
Mcps (mega counts per second), the rate can be increased to 10 Mcps. However, the maximum dark count rate was 1 
Mcps.

In our research, the major objectives are as follows: development of a YAP(Ce)-MPPC detector, decreasing of the dark 
count rate of the detector for photon-counting, development of a simple current-voltage (I-V) amplifier for high-speed 

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detector, an I-V amplifier, an integrator, and a high-speed inverse comparator with a hysteresis circuit. We also carried out CT using gadolinium-based contrast media.

2. METHOD

Figure 1 shows a block diagram of a PC-CT system utilizing the detector. The CT system consists of an X-ray generator, an oscillation linear detector, an integrator with a time constant of 100 ns, an ultra-high-speed comparator (Analog Devices, AD8561), a turntable (Siguma Koki, SGSP-60YAW-OB), a two-stage controller (Siguma Koki, SHOT-602), a counter card (CC; Contec, CNT32-4MT), and a personal computer (PC). The distance between the X-ray source and the YAP(Ce)-MPPC detector (Toreck, YAP-MPPC) is 1.00 m, and the distance from the center of turntable to the scintillator is 55 mm to decrease magnification ratio of an object.

The oscillation-type linear detector is composed of a scan stage (Siguma Koki, SGSP-26-100), the YAP(Ce)-MPPC detector, and the I-V amplifier. X-ray photons are detected using a YAP(Ce) single crystal scintillator and an MPPC (Hamamatsu Photonics, S10362-11-050C), and the photocurrent is amplified by a current-voltage (I-V) amplifier (Fig. 2). In the I-V amplifier, the input voltage $V_{in}$ (V) of an operational amplifier (OA; Analog Devices, AD8032) is given by:

$$V_{in} = 510 I,$$

(1)

where $I$ (A) is the photocurrent through 510 $\Omega$ resistor. The output is then formed into event pulse by the integrator and is sent to the comparator. The comparator (Analog Devices, AD8561) is of an ultra-high-speed type at a single supplied voltage of +5.00 V, and we used an inverse-comparing method with a hysteresis circuit consisting of two resistors for preventing error oscillations from the IC output. A condenser $C_o$ of 1 nF is used to absorb voltage oscillations from a 50 $\Omega$ coaxial cable between the CC and comparator. Logical pulses are produced by the comparator and are counted by the CC and PC. Although both the rise and fall times from the comparator increase, the error oscillation should be prevented corresponding to the circuit for counting logical pulses. The comparator transmits the pulses with heights beyond the lower-level voltage $V_l$ which is regulated using a variable resistor of 10 k$\Omega$ and is measured using a voltage meter.

The 30 ns-decay-time YAP(Ce) scintillator is stuck on the light-receiving surface in the MPPC using epoxy resin and is covered with an aluminum cap with a 0.2 mm-thick aluminum window. The YAP(Ce)-MPPC detector with the amplifier oscillates on the scan stage. The X-ray projection curves for tomography are obtained by repeated linear scans and rotations of the object, and tomograms are reconstructed using the simplest convolution back projection method. Both scan stage and turntable are driven by the two-stage controller. The scan and rotation steps were selected to be 0.5 mm and 1.0°, respectively. In this PC-CT, the scan velocity is 25 mm/s, and X-ray exposure time is 10 min.

Fig. 1. Block diagram of a photon-counting X-ray CT system using a YAP(Ce)-MPPC (multipixel photon counter) detector. The event pulse is generated using an I-V amplifier and an integrator, and the pulse height is dispersed by a high-speed inverse comparator.

Fig. 2. Electric circuit of a device for counting X-ray photons using a YAP(Ce)-MPPC detector. Input voltage $V_{in}$ of an operational amplifier (OA) is produced by the resistor of 510 $\Omega$ and photocurrent $I$ (A) and is equal to 510$I$ (V). Output voltage from the OA is sent to a 100 ns-time-constant integrator.

Proc. of SPIE Vol. 8508 850807-2
3. RESULTS

3.1 X-ray intensity and spectra
The X-ray intensity was measured using an ionization chamber (RAMTEC 1000 plus, Toyo Medic) at 1.0 m from X-ray source and a tube current of 1.0 mA (Fig. 3). At the constant tube current, X-ray intensity increased with increase in tube voltage. At a tube voltage of 80 kV, the X-ray intensity was 70.0 μGy/s.

X-ray spectra used for PC-CT are shown in Fig. 4. Using the detector in conjunction with a multichannel analyzer, the spectra could not be measured owing to an extremely short event-pulse width. Therefore, we used a CdTe detector (XR-100T, Amptek) to measure spectra. Gadolinium K-edge energy of 50.3 keV is shown in the same figure to perform iodine imaging, and X-ray photons with energies just beyond K-edge are absorbed effectively by gadolinium contrast media. At a tube voltage of 80 kV, the peak energy was 33 keV.

3.2 Event pulse and comparator output
Figure 5 shows time relation between the event pulse voltage and the comparator output at a $V_i$ of 0.10 V. In the comparator, because we used an inverse comparing circuit, the initial voltage before the input of event pulse was 3.9 V with a single positive supplied voltage of $+5.00 \text{ V}$. When the event-pulse voltage increased beyond 0.10 V, the comparator voltage decreased to 0 V with a delay time of 10 ns. The voltage increased when the event voltage decreased below 0.10 V. Although both the fall and rise times increased owing to $C_o$, we could prevent the error voltage oscillation.

![Fig. 3. X-ray intensity measured using an ionization chamber placed 1.0 m from the X-ray source at a tube current of 1.0 mA.](image1)

![Fig. 4. X-ray spectra measured by a cadmium telluride (CdTe) detector at a tube voltages of 80 kV. Gadolinium K-edge energy of 50.3 keV is shown in the same figure, and X-ray photons with energies just beyond 50.3 keV are absorbed effectively by gadolinium-based contrast media.](image2)
3.3 Tomography

Tomography was performed at a constant tube voltage of 80 kV and a current of 1.0 mA, and the maximum and minimum densities are denoted as black and white, respectively.

Figure 6 shows tomograms of two glass vials filled with two different density gadolinium media (meeglumine gadopentetate) of 15 and 30 mg/ml, respectively. With increase in $V_t$, the image density difference between two media decreased, and the image granulation increased.

Tomograms of a polymethyl methacrylate (PMMA) phantom with two holes filled with two different density iodine media are shown in Fig. 7. When $V_t$ was increased, the density difference between two holes (media) decreased, and the image granulation also increased owing to low count rates.

The result of tomography of a nude-mouse phantom is shown in Fig. 8. Gadolinium-oxide suspension was used for producing the phantom, and the average particle diameter and density of gadolinium oxide were 700 nm and 20%, respectively. The animal operation was carried out in accordance with the animal experiment guidelines of our university. The suspension of 0.2 ml was injected into the cancerous region of a nude mouse seven days before sacrifice. The phantom was then fixed with formalin, and gadolinium oxide particles in the cancerous region were observed at high contrast at a $V_t$ of 0.10 V. When $V_t$ was increased, the image contrast of particles fell, and the body (muscle) contrast improved.

4. DISCUSSION AND CONCLUSIONS

We developed a simple PC-CT system using a YAP(Ce)-MPPC detector and performed CT using a 25 mm/s-scan linear detector. The maximum count rate was 250 kcps at a $V_t$ of 0.10 V, and the maximum count per measuring point was 5 kc (=250 kc/50). The exposure time for CT was 10 min at a rotation step of 1.0°. The X-ray count rate decreased with increase in $V_t$, and the image granulation also increased.

In tomography, we observed contrast variations with changes in $V_t$ of the comparator at a constant tube voltage of 80 kV, and gadolinium oxide nanoparticles in cancerous region were observed at high contrast with a $V_t$ of 0.10 V. In particular, the image density difference of two gadolinium media in a PMMA phantom at a $V_t$ of 0.10 V was quite large, and the image was almost equal to those obtained by gadolinium K-edge imaging achieved with CdTe detectors.

The spatial resolution is primarily determined by the dimensions of YAP(Ce) crystal for photon counting, and the original resolutions were 1.0×1.0 mm². However, the resolution can be improved with decreases in the scan and rotation steps.
Fig. 6. Tomograms of glass vials filled with two different density gadolinium media of 15 and 30 mg/ml. The image density difference between two media decreased with increase in $V_l$.

Fig. 7. Tomograms of a PMMA phantom with two holes filled with two different density gadolinium media. When $V_l$ was increased, the image density difference between holes (media) decreased, and the image granulation increased.

Fig. 8. Tomograms of a nude mouse phantom with a cancerous region. Gadolinium particles in cancerous region were observed at high contrast. When $V_l$ was increased, the density of the particles increased, and the density of the mouse body (muscle) decreased.

ACKNOWLEDGMENTS
This work was supported by Grants from Keiryo research foundation, Promotion and mutual aid corporation for private schools of Japan, Japan science and technology agency (JST), and Ministry of education, culture, sports, science and technology (MEXT).

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