Investigation of X-ray photon-counting using ceramic-substrate silicon diode and its application to gadolinium imaging

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Abstract
X-ray photon counting was performed using a silicon X-ray diode (Si-XD) at tube voltages ranging from 20 to 100 kV. The Si-XD is a high-sensitivity Si photodiode selected for detecting X-ray photons, and the photons are directly counted using the Si-XD. Count rates of 10 kilo-counts per second (kbps) are achieved, and the photon count rate has been increased to 15 Mcps/mm² using a 1 × 1 × 1 mm³-cube zinc oxide (ZnO) crystal. However, it is difficult to disperse X-ray photon energy using the MCA owing to 1-Mcps dark counting.

1. Introduction
Monochromatic parallel X-ray beams are produced using a synchrotron and silicon single crystals, and iodine K-edge angiography is performed to observe ne coronary arteries at high contrast. Using this camera, 200-µm-diameter fine arteries are observed at high contrast by selecting photons with energies just beyond 33.2 keV.

The K-edge angiography can also be carried out utilizing the photon energy dispersion, and we have developed an energy-dispersive (ED) X-ray camera using a cadmium telluride (CdTe) detector and an x–y stage. Using this camera, 200-µm-diameter fine arteries are observed at high contrast by selecting photons with energies just beyond 33.2 keV.

The energy-dispersive X-ray computed tomography (ED-CT) system is a photon counting system (PC-CT) system with a high energy resolution of approximately 1% at 122 keV, and the K-edge imaging can be performed. On the other hand, we define the PC-CT has a low-energy-resolution detector, and it is difficult to measure X-ray spectra using the detector in the PC-CT.

To perform ED-CT, a preclinical CT system has been developed using a CdTe array, and gadolinium (Gd) K-edge imaging has been performed. Successively, several ED-CT systems have been developed using an oscillation-type linear CdTe scan detector to carry out iodine and Gd K-edge imaging. In the early developed ED-CT, the photon energy range was selected using a multichannel analyzer (MCA). Subsequently, in the recent ED-CT, the energy dispersion is carried out using a comparator device to increase the photon count rate by determining the discrimination voltage ($V_1$) of the event pulse, because the maximum count rate of the MCA is 10 kilo-counts per second (kcps). In this case, the $V_1$ is proportional to the minimum photon energy ($E_{\text{min}}$) for X-ray imaging.

Recently, photon-counting X-ray CT (PC-CT) systems have been developed to reduce the X-ray exposure time for CT by increasing photon-count rate. These systems also employ a linear scanning detector with a multipixel photon counter (MPPC) module and short-decay time scintillators. In particular, the photon count rate has been increased to 15 Mcps/mm² using a 1 × 1 × 1 mm³-cube zinc oxide (ZnO) crystal. However, it is difficult to disperse X-ray photon energy using the MCA owing to 1-Mcps dark counting.

Most X-ray CT systems utilize detectors consisting of silicon photodiodes (Si-PD) and long-decay-time scintillators. Without a scintillator crystal, a high-sensitivity Si X-ray diode (Si-XD) has been found, and we have developed a high-sensitivity X-ray CT system using a linear scanning detector with the Si-XD. In addition, because the sensitivity was roughly proportional to the tube current from 1 to 100 mA at a constant tube voltage, the semiconductor dosimeter for measuring high-dose-rate pulsed X-rays in medical radiography has also been developed using the Si-XD.

The reasons to perform experiments in the direct X-ray counting using Si-XD detectors are as follows: to simplify the X-ray detector system without scintillators, to realize a low-cost detector system for CT, to reduce the bias voltage of the detector to below 10 V, and to improve the spatial resolution of the detector.

Because it is easy to count X-ray photons using the Si-XD, the photon counting characteristics should be measured to perform K-edge imaging and dual-energy subtraction. In addition, although the detection efficiency of Si substantially decreases with increase in the photon energy, Gd-K-edge imaging would be carried out using photons with energies beyond K-edge energy of 50.3 keV. The expected benefits of the photon counting utilizing a ceramic-substrate Si-XD as follows: to detect scattering X-ray photons from the ceramic substrate, to increase detection efficiency, and to control image contrast by selecting the $E_{\text{min}}$. In our research, major objectives are as follows: to count X-ray photons using a Si-XD, to measure the event-pulse-height (EPH) spectra using an MCA, to detect high-energy photons for imaging, and to perform PC-CT using Gd media and a comparator. Therefore, we investigated X-ray photon counting using a ceramic-substrate Si-XD and performed the Gd-K-edge imaging.
Fig. 1. (Color online) Block diagram for counting X-ray photons using a Si-XD. Photons are counted directly using the Si-XD without a scintillator. The EPH spectra were measured using an MCA, and the structure of the Si-XD is shown in the same figure.

2. Experimental methods

2.1 Photon-counting using Si-XD

Figure 1 shows a block diagram for counting X-ray photons using a Si-XD. The X-ray generator (R-tec RXG-0152) has been developed to produce low-dose-rate X-ray beams by decreasing tube current ranging from 1.0 µA to 2.0 mA. The total filtration of the X-ray tube unit is 2.5 mm in aluminum (Al) equivalent thickness. The X-ray photons from an X-ray tube are detected by the Si-XD, and the photocurrents flowing through the Si-XD are amplified by the charge sensitive (Clear Pulse Model 580) and shaping (Clear Pulse Model 4419) amplifiers. The event pulses from the shaping amplifier are sent to an MCA (PECT MCA4000) to measure EPH spectra with changes in the tube voltage. In the Si-XD, a ceramic-type Si photodiode (Hamamatsu S1087-01) is covered with an Al cap with a 0.2-mm-thick Al window, and the dimensions of the light receiving surface are $1.3 \times 1.3$ mm². Successively, the Si-XD is shielded using both an Al case with a 25-µm-thick Al window and a Bayonet Neill Concelman (BNC) connector, and the bias voltage of the Si-XD was 10 V. The distance between the target in the X-ray tube and the shielded Si-XD was 1.00 m.

2.2 PC-CT system

A PC-CT system is used for observing image contrast variations with changes in the minimum photon energy, and a block diagram of the PC-CT system utilizing the detector is shown in Fig. 2. The CT system consists of an X-ray generator, a 25.0-mm/s linear X-ray scanner, a turntable (Siguma Koki SGSP-60YAW-OB), a two-stage controller (Siguma Koki SHOT-602), two amplifiers, a comparator device, a frequency-voltage converter (FVC; Toreck FDC13), an analog-digital converter (ADC; Contec AI-1608AY-USB), 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 turntable to the shielded Si-XD tip is 40 mm to decrease the magnification ratio of an object. The oscillation-type X-ray scanner is composed of a scan stage (Siguma Koki SGSP-26-100), the shielded Si-XD, and the charge-sensitive amplifier. The PC-CT is accomplished by repeated linear scans and rotations of the object, and projection curves of the object are obtained by the linear scanning at a tube current of 2.0 mA. The scanning is conducted in both directions of its movement, and tomograms are reconstructed using the simplest convolution back projection method.

X-ray photons are detected using the Si-XD, and the photocurrent is amplified using the two amplifiers. Then, the event pulses are sent to the comparator for determining $V_l$, and the logical pulses are converted into smoothed output voltages using a frequency-voltage converter (FVC; Toreck FVC13A). Then, the voltages are sent to the PC through the ADC. Using the developed FVC, the output voltage is proportional to the count rate.

The circuit diagram of a comparator device is shown in Fig. 3. A comparator chip (STMicroelectronics TS3022) is used for determining $V_l$, which is controlled by a variable resistor of 10 kΩ and is measured using direct-current (DC) voltage meter. We used an inverse-comparing method with a hysteresis circuit consisting of two resistors for preventing error oscillations from the IC output. Using this comparator device, initial comparator output before inputting event signal...
is 5.0 V, and the initial comparator output $V_c(t)$ as a function of time $t$ decreases to 0 V when event-signal voltage $V_e(t)$ increases beyond $V_l$. Then, $V_c(t)$ increased from 0 to 5.0 V when $V_e(t)$ decreased below $V_l$.

### 3. Results

#### 3.1 X-ray dose rate and spectra

The measurement of X-ray dose rate is very important for inferring the skin dose for objects. The X-ray dose rate from the X-ray generator was measured using an ionization chamber (Toyo Medic RAMTEC 1000 plus) placed 1.0 m from the X-ray source (Fig. 4). At a constant tube current of 2.0 mA, the X-ray dose rate increased with increasing tube voltage. The X-ray dose rates at tube voltages of 50 and 100 kV were 61.3 and 222 µGy/s, respectively.

X-ray spectra used for CT are shown in Fig. 5. In order to measure X-ray spectra, we used a CdTe detector (Amptek XR-100T). Gd K-edge energy (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 atoms.

#### 3.2 EPH spectra

The EPH spectra obtained using the Si-XD is shown in Fig. 6. The photon energy is determined using two-point calibration by measuring the EPH spectra at tube voltages of 20 to 100 kV. In the calibration, we utilized logarithmic spectra with a minimum photon count of 10 to determine the maximum energy without pileups of the event pulse. Because it is unable to regulate the tube voltage to 20 kV using the X-ray generator in the CT system, we used a soft X-ray generator (R-tek RIX-20) with a 0.5-mm-thick beryllium window with tube voltages ranging from 15 to 100 kV. In the EPH spectra, the maximum photon energy corresponded to the tube voltage. At a tube voltage of 100 kV, we could not observe any K-series characteristic X-rays from a tungsten target in the tube. In other results, the maximum photon energy seldom varied by the filtration at a constant tube voltage. In addition, the photon count substantially increased with decreasing photon energy. Therefore, X-ray photons were detected directly by the Si substrate, and the scattering photons from the ceramic were also detected as low-energy photons. Thus, the $E_{\text{min}}$ could be regulated by the $V_l$.

#### 3.3 Comparator output

Figure 7 shows time relationship between event-pulse voltage and comparator output at a $V_l$ of 1.2 V. The pulse height and the width were 2.3 V and 3 µs, respectively. The initial voltage of the comparator output before the input...
of event pulse was 5.0 V, and the comparator output $V_c(t)$ substantially decreased to 0 V when event-pulse voltage $V_e(t)$ increased beyond 1.2 V with a delay time of 33 ns. Successively, $V_c(t)$ increased from 0 to 5.0 V when $V_e(t)$ decreased below 1.2 V.

The minimum discrimination voltage $V_1$ (V) of the event-pulse height as a function of the minimum photon energy $E_{\text{min}}$ (keV) is written by

$$V_1 = \frac{E_{\text{min}}}{39.2} + 0.76.$$  \hspace{1cm} (1)

At a $V_1$ of 1.2 V, the minimum discrimination photon energy $E_{\text{min}}$ is calculated to be 17 keV using Eq. (1).

### 3.4 Tomography

Tomography was performed at a scan step of 0.5 mm and a rotation step of 1.0°, and the maximum and minimum densities are denoted as black and white, respectively. On the other hand, tomograms are obtained as JPEG files, and the maximum and minimum densities are defined as white and black, respectively.

Figure 8 shows tomograms of two 15-mm-diam glass vials filled with Gd media of two different densities (meeglumine gadopentetate), 15 and 30 mg/ml, respectively. At a tube voltage of 100 kV, the image densities of the two media increased with increasing $E_{\text{min}}$ from 10 to 50 keV. The image densities of the two media slightly increased with decreases in the tube voltage from 100 to 50 kV at an $E_{\text{min}}$ of 10 keV.

The density analysis of the two glass vials in Fig. 8 with a tube voltage of 100 kV is shown in Fig. 9. At an $E_{\text{min}}$ range of 10–100 keV, the image densities of the two glass vials were higher than those of the two media. When the $E_{\text{min}}$ was increased from 10 to 50 kV, the image densities of the two media increased.

Figure 10 shows the result of the tomography of a rabbit-head phantom. The phantom was made from a real rabbit head, and the blood vessels were filled with gadolinium oxide (Gd$_2$O$_3$) microparticles. The animal operation was carried out in accordance with the animal experiment guidelines of our university. Radiography (angiography) was performed for reference with a flat-panel detector (FPD; Rad-icon Imaging 1024 EV) to observe blood vessels. In radiography, fine blood vessels were observed because the pixel sizes are $48 \times 48$ μm$^2$. In tomography, when the $E_{\text{min}}$ was increased from 10 to 50 keV at a tube voltage of 100 kV, the image densities of the bones decreased, and the image contrast of the vessels improved. In particular, the image contrast of the lower jaw fell with increasing $E_{\text{min}}$. On the other hand, the image contrast of thick vessels substantially fell with decreasing tube voltage from 100 to 50 kV at an $E_{\text{min}}$ of 10 keV.
4. Discussion and conclusions

We performed the X-ray photon counting and measured the EPH spectra using a ceramic-substrate Si-XD and an MCA. In the EPH spectra, the photon count substantially increased with decreasing photon energy. Therefore, penetrating X-ray photons through the Si-XD are scattered and absorbed by the ceramic substrate behind Si-XD, and the ceramic produced fluorescent photons. Then, the Si-XD detected low-energy photons from the substrate.

In the PC-CT system, the image contrast of Gd media with decreases in the rate substantially increased to beyond 100 kcps/ pixel (≈ 42.5 kc/50). Furthermore, the rate substantially increased to beyond 100 kcps/pixel with decreases in $E_{\text{min}}$. Therefore, a PC-CT system in the 2 mA–150 kV range in biomedical imaging might be realized without scintillators in the near future.

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To perform photon counting using the Si-XD, we used ready-made charge sensitive and shaping amplifiers. However, it is easy to develop a simple high-speed amplifier system for the Si-XD using a general-purpose operational amplifier. In addition, a Si-PIN diode with a cutoff frequency of 100–500 MHz would be useful for high-count-rate energy dispersion for PC-CT systems.

In this PC-CT with a tube voltage of 100 kV, a current of 2.0 mA, and an $E_{\text{min}}$ of 10 keV, the maximum count rate was 42.5 kcp/pixel, and the photon count per measuring point was calculated as 0.85 kc/pixel ($=42.5$kc/50). Furthermore, the rate substantially increased to beyond 100 kcp/pixel with decreases in $E_{\text{min}}$.

Fig. 10. (Color online) Tomography of a rabbit-head phantom. The blood vessels were filled with Gd$_2$O$_3$ microparticles. When the $E_{\text{min}}$ was increased from 10 to 50 keV at a tube voltage of 100 kV, the image density of the bones decreased, and the vessel contrast slightly improved. With decreasing tube voltage from 100 to 50 kV at an $E_{\text{min}}$ of 10 keV, the image contrast of the vessels substantially fell.

The pixel sizes for reconstructing CT image were equal to the scan steps and had values of 0.5 $\times$ 0.5 mm$^2$. To improve the spatial resolution, the dimensions should be reduced.

Although there are a few readily available FVCs, the output voltage was not proportional to the count rate. Therefore, to avoid this, we used the FVC with a microcomputer and an integrator to improve the image granulation of tomograms. The microcomputer produced constant-width logical pulses, and the FVC output voltage was in proportion to the rate.