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High-sensitivity X-ray computed tomography system using a direct-conversion silicon-PIN diode and a 50-ms-time-constant integrator

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Abstract

The silicon-PIN X-ray diode (Si-PIN-XD) is a high-sensitivity Si-PIN photodiode selected for detecting X-rays. X-ray photons are directly detected using the Si-PIN-XD without a scintillator, and the photocurrents from the diode are converted into voltages and amplified using current-voltage (I-V) and voltage-voltage (V-V) amplifiers. The output voltage from the V-V amplifier is sent to an analog-digital converter through a 50-ms-time-constant integrator to compensate for the image granulation. Computed tomography (CT) was accomplished by repeated linear scans and rotations of an object, and projection curves of the object are obtained by the linear scan. 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 °. The maximum output voltage from the V-V amplifier is regulated to 3.0 V to perform CT, and the CT was carried out using iodine-based contrast media and bremsstrahlung X-ray spectra.

Keywords: Si-PIN X-ray diode, direct-conversion imaging, scintillator-less, sensitivity measurement, X-ray CT, iodine imaging

1. Introduction

Recently, several photon-counting energy-dispersive computed tomography (ED-CT) systems[1-6] have been developed to perform K-edge imaging using iodine (I) and gadolinium (Gd) media. To carry out K-edge imaging, monochromatic X-ray photons with energies beyond the K-edge energy were used. In these systems, we usually used cadmium telluride (CdTe) detectors with an energy resolution of 1 % at 122 keV, and blood vessels were observed at high contrast.

The K-edge imaging can also be performed utilizing monochromatic bremsstrahlung X-rays by the

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filtration, and I-K-edge imaging has been performed using a direct-conversion silicon X-ray diode (Si-XD)[7, 8] with a ceramic substrate. Subsequently, a Si-PIN-XD[9] is also useful for measuring X-ray spectra, and we are very interested in performing I-K-edge imaging utilizing aluminum filtration.

In our research, major objectives are as follows: to develop a direct-conversion low-cost detector for CT without a scintillator, to load an X-ray detecting module, to improve image granulation using an integrator, and to perform I-K-edge CT. Therefore, we constructed a high-sensitivity X-ray CT system using a Si-PIN-XD and an integrator and performed I-K-edge imaging.

2. Experimental methods

2.1. Sensitivity of Si-PIN-XD

Figure 1 shows the experimental setup for detecting low-dose-rate X-rays. The Si-PIN-XD (S5971, Hamamatsu) is shielded using an aluminum (Al) case with a 25 μm-thick Al window and a BNC connector. In the Si-PIN-XD detector, the X-rays are directly detected without a scintillator. The photocurrents flowing through the Si-PIN-XD are converted into voltages and amplified using current-voltage (I-V) and voltage-voltage (V-V) amplifiers in the X-ray detecting module, and the output

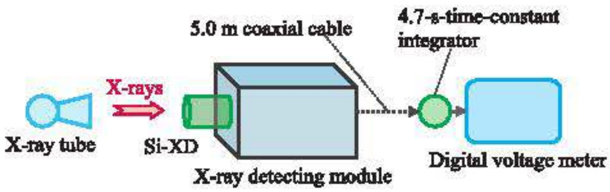

Fig. 1. Block diagram for measuring low-dose-rate X-rays using a Si-PIN-XD and an X-ray detecting module. Si-PIN-XD is connected directly to the module.

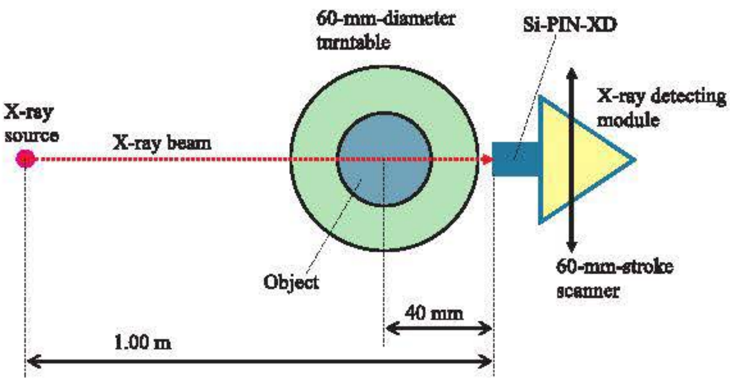

Fig. 2. Experimental setup for performing tomography using a 60-mm-stroke X-ray scanner with the Si-PIN-XD.

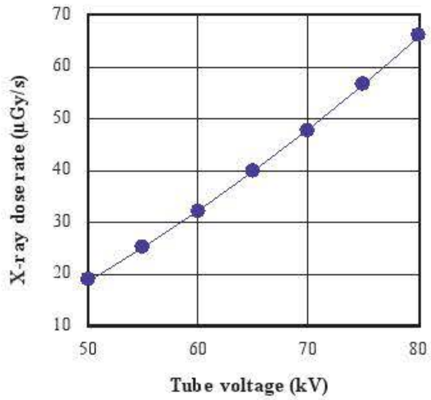

Fig. 3. X-ray dose rate measured using an ionization chamber placed 1.0 m from the X-ray source with a 3.0-mm-thick Al filter and a tube current of 2.0 mA.

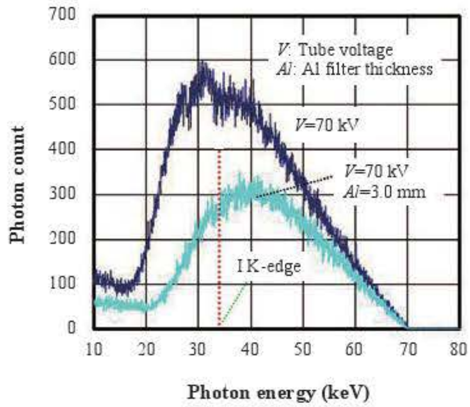

Fig. 4. X-ray spectra measured using a CdTe detector at a tube voltage of 70 kV. I-K-edge energy of 33.2 keV is shown in the same figure for reference, and X-ray photons with energies just beyond 33.2 keV are absorbed effectively by I-based contrast media.

voltages are measured using a digital voltage meter and a 4.7-s-time-constant integrator for smoothing voltages.

2.2. X-ray CT System

A block diagram of an X-ray CT system utilizing the detector and the module is shown in Fig. 2. The CT system consists of an X-ray generator (RXG-0152, R-tec), a 25.0-mm/s-scan linear X-ray detector, a turntable (SGSP-60YAW-OB, Siguma Koki), a two-stage controller (SHOT-602, Siguma Koki), 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 detector is 40 mm to decrease the magnification ratio of an object. The oscillation-type linear detector is composed of a scan stage (SGSP-26-100, Siguma Koki) , the Si-PIN-XD, and the X-ray detecting module. The output voltage from the module is sent to an analog-digital converter (ADC) through a 50-ms integrator.

The Si-PIN-XD with the module oscillates on the scan stage with a velocity of 25 mm/s. The X-ray projection curves for tomography are obtained by repeated linear scans and rotations of the object. Both the scan stage and turntable are driven by the two-stage controller. In this X-ray CT, the X-ray exposure time for obtaining a tomogram is 10 min with a scan step of 0.5 mm and a rotation step of 1 °.

3. Results

3.1. X-ray dose rate and spectra

The measurement of X-ray dose rate is very important for inferring the absorbed dose in objects.

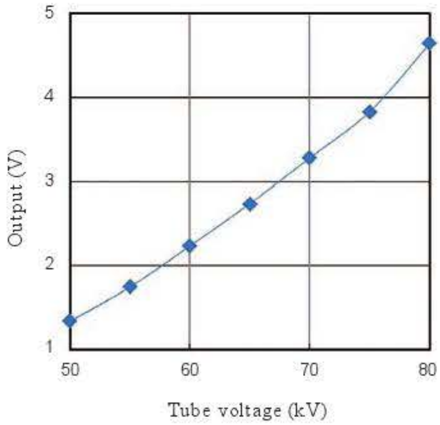

Fig. 5. Output voltages from the module with changes in the tube voltage using the Al filter at a tube current of 2.0 mA.

The X-ray dose rate from an X-ray generator was measured using an ionization chamber (RAMTEC 1000 plus Toyo Medic) and a 3.0-mm-thick aluminum (Al) filter at a tube current of 2.0 mA (Fig. 5). The chamber was placed 1.0 m from the X-ray source. At a constant tube current, X-ray dose rate increased with increasing tube voltage. At a tube voltage of 70 kV, the X-ray dose rate was 48.2 μGy/s.

X-ray spectra used for CT are shown in Fig. 4. To measure X-ray spectra, we used a CdTe detector (XR-100T, Amptek). I-K-edge energy (33.2 keV) is shown in the same figure for reference, and X-ray photons with energies just beyond the K-edge are absorbed effectively by I atoms. The maximum photon energy corresponded to the tube voltage of 70 kV, and the bremsstrahlung peak energy was approximately 40 keV by the insertion of the Al filter. Therefore, it is possible to perform the I-K-edge CT, since the peak energy was beyond the K-edge.

3.2. Output voltage

Figure 5 shows the output voltages from the Si-PIN-XD measured using the X-ray detecting module. At a tube current of 2.0 mA using the Al filter, the output increased with increasing tube voltage, and the output voltage was 3.3 V with a tube voltage of 70 kV. Thus, it is very easy to employ the Si-PIN-XD as a detector in a high-sensitivity low-dose-rate X-ray CT system.

3.3. Tomography

Tomography was performed using bremsstrahlung photons as in Fig. 4 at a constant tube voltage of 70 kV, and the tube currents with and without filtration were 2.0 and 1.0 mA, respectively. In tomography, the maximum and minimum densities are denoted in 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.

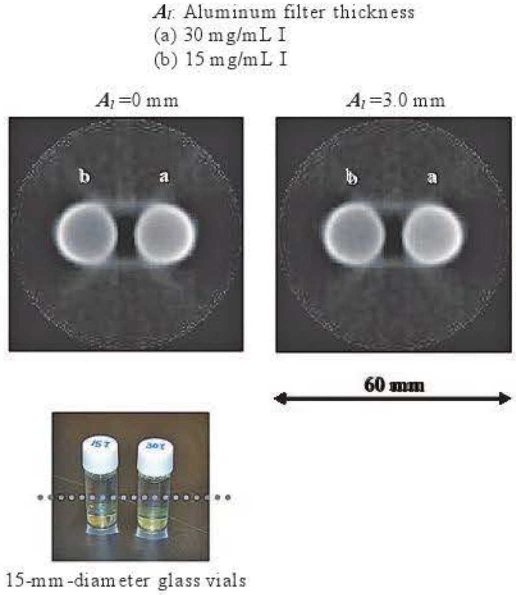

Fig. 6. Tomograms of glass vials filled with two different density I media of 15 and 30 mg/mL. The image densities of the two media slightly increased with the insertion of the Al filter.

Tomograms of two glass vials filled with I media (iopamidol) of two different densities 15 and 30 mg/mL are shown in Fig. 6. At a constant maximum energy of 70 keV, the image contrast of the two vials improved slightly with the Al filtration.

Figure 7 shows image density analysis of the two tomograms in Fig. 6 using a program Image J. By the Al filtration, the image densities of the two media slightly increased, since bremsstrahlung peak energy shifted to beyond the I-K-edge.

The result of the tomography of a dog-heart phantom is shown in Fig. 8. Coronary arteries are filled with I-based microspheres 15 μm in diameter. The animal operation was carried out in accordance with the animal experiment guidelines of our university. When the Al filter was used, the image density of muscle slightly decreased, and the image contrast of the blood vessels were slightly improved.

4. Discussion and conclusions

We measured the relative sensitivity of a direct-conversion Si-PIN-XD and developed a high-sensitivity X-ray CT system. Subsequently, we performed I-K-edge CT using a 25 mm/s-scan linear detector, I contrast media, and a 3.0 mm-thick Al filter. To perform I-K-edge CT, X-ray photons just beyond 33.2 keV are effective. Thus, the Al filter is useful for carrying out I-K-edge imaging by absorbing low-energy photons, since the bremsstrahlung peak energy was 40 keV at a tube voltage of 70 kV.

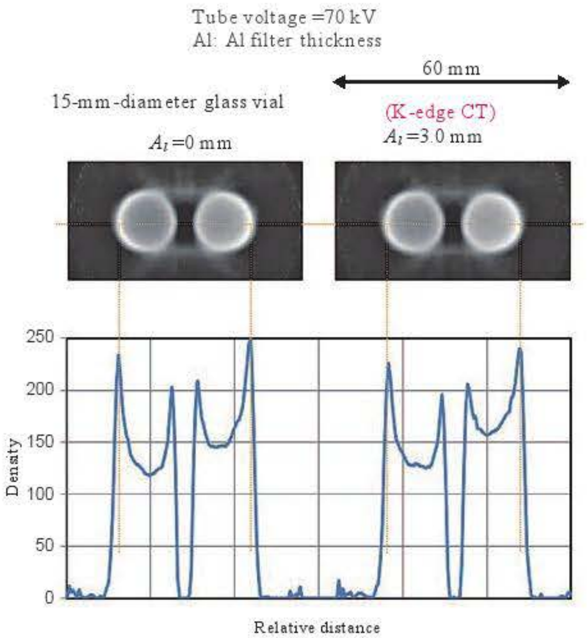

Fig. 7. Image density analysis of the two glass vials using a program (Image J).

The spatial resolution is primarily determined by the dimensions of the X-ray-receiving surface of 1.2 mm in diameter, and the original resolutions were $1.2\times1.2\ \mathrm{mm}^2$. To improve the spatial resolution, the dimensions should be reduced, and a lead pinhole would be useful.

We also found a high-sensitivity Si-XD, and the sensitivity of the Si-XD is higher than that of Si-PIN-XD. Therefore, a scintillator-less X-ray detector system for CT might be realized by removing scintillators from the conventional CT detectors.

Acknowledgments

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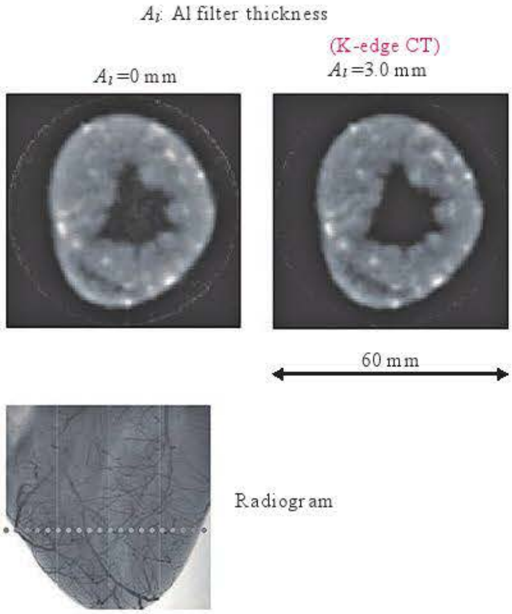

Fig. 8. Tomograms of a dog-heart phantom. Coronary arteries were filled with I-based microspheres, and thick arteries were observed at high contrast. Radiography (angiography) was performed with a flat-panel detector (FPD; Rad-icon Imaging 1024EV) to observe blood vessels.

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