Image-quality improvement in pileup-less cadmium-telluride X-ray computed tomography using a frequency-voltage converter and its application to iodine imaging

Anna SHIMAMURA¹, Eiichi SATO¹, Shunsuke SHIKANAI¹, Kei KITAKAMI¹, Ippeki NAKAYA¹, Wakano NISHIMURA¹, Yuich SATO¹, Satoshi YAMAGUCHI¹,²,³, Yasuyuki ODA¹, Osahiko HAGIWARA¹, Hiroshi MATSUKIYO¹,²,³, Toshiyuki ENOMOTO¹,²,³, Manabu WATANABE¹,²,³, Shinya KUSACHI¹,²,³, Shigeru EHARA¹,²,³

¹Department of Physics, Iwate Medical University,
2-1-1 Nishitokuta, Yahaba, Iwate 028-3694, Japan
²Central Radiation Department, Iwate Medical University Hospital,
19-1 Uchimaru, Morioka, Iwate 020-0023, Japan
³Department of Radiology, School of Medicine, Iwate Medical University,
19-1 Uchimaru, Morioka, Iwate 020-0023, Japan

Abstract: To perform investigation of image-quality improvement in photon-counting energy-dispersive X-ray computed tomography (ED-CT) under low count rates, we have developed an ED-CT system using a frequency-voltage converter (FVC) and a cadmium telluride detector. The FVC is used to improve the image granularity by compensating for the photon-count fluctuation. In the ED-CT, the photon energy range is regulated using a multichannel analyzer (MCA) by determining both the maximum and minimum energies. The FVC consists of a microcomputer, an integrator, and an operational amplifier. When 0.2-µs-width event logical pulses from the MCA are sent to the microcomputer, 5-µs-width 5.0 -V-height logical pulses are produced from the microcomputer and sent to the integrator to produce the time-average voltage which is proportional to the count rate. The integrator output is then amplified by the amplifier, and the FVC output is input to an analog-digital converter. Pileup-less CT was accomplished by repeated linear scans and rotation of the object at a tube voltage of 70 kV and a count rate of 1.9 kilo-counts per second or below. We also performed the iodine K-edge CT and the energy subtraction.

Keywords: X-ray CT, CdTe, pileup-less, frequency-voltage converter, integrator

1. INTRODUCTION

To measure X-ray spectra, we usually use cadmium telluride (CdTe) detectors, and we have developed several energy dispersive (ED) imaging systems using a cooled CdTe detector with an energy resolution of 1% at 122 keV. Using a penetration type ED camera [1], we carried out iodine [1] K-edge angiography with energies just beyond I-K-edge energy 33.2 keV, and 200-µm-diameter blood vessels were clearly observed. Successively, several energy dispersive computed tomography (ED-CT) systems [2-6] have been developed to perform K-edge CT using I and gadolinium (Gd) media. In addition, a preclinical ED-CT system [7,8] has been developed using a CdTe array to perform Gd K-edge imaging. The ED-CT system is of a photon counting CT (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 that the PC-CT [9,10] has a low-energy-resolution detector, and it is difficult to measure X-ray spectra using the detector in the PC-CT.

Recently, a PC-CT system has been developed using detectors consisting of a multipixel photon counter (MPPC) module and a short-decay-time scintillator to increase the count rate. Using this PC-CT, the rate has been increased to 15 mega-counts per second (Mcps) [10] using a 1-ns-decay-time zinc oxide crystal, and the image granularity was slightly improved. However, the granularity increased with decreasing count rate. In addition, it is unable to obtain effective image-contrast variations with changes in the discrimination voltage of the event pulse using a comparator; the comparator generates logical pulses when the event pulse voltage increases beyond the discrimination voltage.

It is well known that the image quality of tomograms improves with increasing photon count per measuring point by increasing the photon count rate at a constant exposure time for CT. On the other hand, the count rate fluctuates at a low rate region below 10 kcps, and the count rate fluctuation makes image granulations in tomography.

To prevent piling of the event pulse from a conventional CdTe detector system made by Amptek, the maximum rate is approximately 5 kcps. If we assume that the scan velocity is 25 mm/s using a single detector in the first-generation CT system, the photon count per measuring point with a scan step of 1 mm is calculated as 200 counts using whole photons. In addition, the photon count decreases by selecting the energy range, and the image granularity of tomography substantially increases below 100 counts. A computer program helps smoothing the count rate, but also the smoothing can be carried out using a frequency-voltage converter (FVC) with an integrator.

In our research, major objectives are as follows: to load a developed FVC for compensating for the photo-count...
fluctuations, to keep the energy resolution of 1% at 122 keV by preventing pileups of the event pulse, to perform ED-CT by determining the energy range using a multichannel analyzer (MCA) for the pulse height analysis, and to carry out imaging. Therefore, we constructed an ED-CT system with an FVC operated at a tube current of 24 μA and performed K-edge CT and energy subtraction at a maximum count rate of 1.9 kcps.

2. EXPERIMENTAL SETUP

Fig. 1 shows the experimental setup of the main components in an ED-CT system. The ED-CT system is of a first-generation type [5] and consists of an X-ray generator (R-tec, RXG-0152), a turntable (Siguma Koki, SGSP-60 YAW-OB), a scan stage (Siguma Koki, SGSP-26-100), a two-stage controller (Siguma Koki, SHOT-602), a CdTe detector system with charge-sensitive and shaping amplifiers (Amptek, XR-100T), an MCA (γPGT, MCA4000), an FVC, an analog-digital converter (ADC; Contec, AI-1608AY-USB), and a personal computer (PC). The distance between the X-ray source and the CdTe detector is 1.00 m, and the distance from the center of turntable to the detector is 40 mm to decrease magnification ratio of an object. A 0.7-mm-diameter lead pinhole is set just in front of the CdTe detector to improve the spatial resolution. The CdTe detector with the charge-sensitive amplifier oscillates on the scan stage with a maximum velocity of 25 mm/s and a stroke of 60 mm. 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. Two step values of the linear scan and rotation are selected to be 0.5 mm and 1.0°, respectively.

Fig. 1 Experimental setup of the main components in an ED-CT system.

The block diagram for counting X-ray photons using a CdTe detector and an FVC is shown in Fig. 2. The X-ray photons are detected by the CdTe detector to perform photoelectric conversion, and the photocurrents are converted into voltages and are amplified using charge-sensitive and shaping amplifiers. The event pulses from the shaping amplifier are sent to the MCA for determining the energy range using the PC, and logical event pulses from the MCA are input to the FVC for smoothing the photon count rate. Currently, the MCA is used to carry out pulse height analysis for measuring X-ray spectra and to count photons in the selected energy range by determining both the maximum and minimum photon energies. During the ED-CT, both the selected X-ray spectra and the count rate are displayed on the PC. The FVC output is input to the ADC, and tomograms are reconstructed by the PC.

The simplified circuit diagram of the FVC is shown in Fig. 3. Although there are a few readymade FVC chips (New Japan Radio NJM4151 and Analog Devices AD650), we constructed an FVC using a microcomputer (Atmel, ATMEGA 168P-20 PU) and an operational amplifier (Texas Instruments, LMC 662). When the 0.2-μs-width logical pulses from the MCA are sent to a microcomputer, the 5-μs-width 5.0-V-height logical pulses are produced by the microcomputer and are input to an integrator to generate time-average voltage of the microcomputer output. The pulse width can be controlled by the microcomputer, and the integrator output increases with increasing width. In this case, the integrator output $V_i$ is written by:

$$V_i = hfw$$

where $h$ is the pulse height (5.0 V), $f$ is the count rate (frequency), and $w$ is the pulse width (5 μs). If we assume that $f$ is 2.0 kcps, $V_i$ is calculated to be 0.05 V.

The smoothed integrator output is amplified using an operational amplifier chip. The time constant of the integrator is controlled using a variable resistor $R_i$, and the maximum output voltage is regulated to below 5.0 V using a variable resistor $R_v$.  

Fig. 2 Block diagram for counting X-ray photons using a CdTe detector and an FVC.

Fig. 3 Circuit diagram of the FVC for improving the image granulation.
3. RESULTS

3.1 X-ray Dose Rate and Spectra

The measurement of X-ray dose rate is quite important for inferring the skin dose for objects. The X-ray dose rate from an X-ray generator was measured using an ionization chamber (Toyo Medic, RAMTEC 1000 plus) at a tube current of 24 μA without filtration. The chamber was placed 1.0 m from the X-ray source. At a constant tube current, the X-ray dose rate increased with increasing tube voltage (Fig. 4). At a tube voltage of 70 kV, the X-ray dose rate was 1.34 μGy/s.

X-ray spectra used for CT are shown in Fig. 5. To measure X-ray spectra, we used the CdTe detector in the ED-CT system. In the whole spectra [Fig. 5(a)] with energies ranging from 20 to 70 keV, the count rate was 4.6 kcps. I-K-edge energy (33.2 keV) is also shown in the same figure for reference. Next, the X-ray photons with energies ranging from 20 to 33 keV are useless for imaging I media [Fig. 5(b)], and the count rate had a value of 1.5 kcps. The X-ray photons from 34 to 50 keV are useful for carrying out I-K-edge CT with a count rate of 1.9 kcps [Fig. 5(c)].

3.2 Electric Characteristics in the FVC

To measure the output voltages, we used a digital oscilloscope (Tektronix, TDS2012C). Fig. 6 shows the time relationship among the logical pulses from the MCA, the logical pulses from the microcomputer, and the FVC output. The logical pulses from the MCA and the microcomputer were measured simultaneously. The microcomputer output was produced with a delay of 7.5 μs from the input of the MCA pulse. In the I-K-edge CT, the maximum output voltage from the FVC was approximately 2 V corresponding to 1.9 kcps.

3.3 Tomography

Tomography was performed at a constant tube voltage of 70 kV, a tube current of 24 μA, and the reconstructed maximum and minimum relative photon counts 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.

Tomograms of two glass vials filled with I media (iopamidol) of two different densities 15 and 30 mg/ml are shown in Fig. 7. Compared with a tomogram using photons below 33 keV, the image density difference between two media was large utilizing I-K-edge CT. The K-edge energy-subtraction tomogram $T_s(x,y)$ was calculated using an equation

$$T_s(x,y) = T_l(x,y) - 0.75 T_h(x,y)$$

where $T_l(x,y)$ is a tomogram utilizing I-K-edge CT, $T_h(x,y)$ is a tomogram using low-energy photons ranging from 20 to 33 keV, and $x$ and $y$ are coordinates. Although the image granulation was observed in the energy-subtraction tomogram

---

Vol.31 No.2 (2014) — 37 —
owing to low count rates by the subtraction, the image density difference between the two media was large. In addition, image density difference between the two media was quite large using I-K-edge CT after the density control; the back ground density was normalized to the minimum \( \# \text{black} \).

The result of the tomography of a dog-heart phantom is shown in Fig. 8. Coronary arteries are filled with I-based microspheres of 15 \( \mu \text{m} \) in diameter. The animal operation was carried out in accordance with the animal experiment guidelines of our university. Using I-K-edge CT, the image density of muscle decreased, and the image contrast of coronary arteries was high. In addition, only the thick arteries were visible utilizing the energy subtraction and the I-K-edge CT after the density control. In the subtraction image, white noises were caused by the increases in the statistical errors.

4. DISCUSSION AND CONCLUSIONS

We performed investigation of image-quality improvement in pileup-less ED-CT system using an FVC and a CdTe scanner. Because the energy resolution of the readily available CdTe detector in the ED-CT system was 1\% at 122 keV, the energy resolution was kept to the same value under the pileup-less condition. In this case, the tube current was reduced to 24 \( \mu \text{A} \) to prevent pileups of the event pulse.

At count rates of 1.9 kcps or below, we also carried out the I-K-edge CT and the energy subtraction. Therefore, I media can be observed at high contrast. Fig. 9 shows tomograms of two glass vials and a dog heart as in Figs. 7 and 8, respectively. These images were reconstructed using a counter card (Contec, CNT32-4MT) without the FVC. Utilizing I-K-edge CT, although

Fig. 7 Tomograms of two glass vials filled with I media of two different densities of 15 and 30 mg/ml. Using I-K-edge CT, the image density difference between two media increased, and image density difference between the two media was quite large using I-K-edge energy subtraction. After the density control, the density difference was also large utilizing I-K-edge CT.

Fig. 8 Tomograms of a dog-heart phantom. Coronary arteries were filled with I-based microspheres, and thick arteries were observed at high contrast using the I-K-edge CT. Only the thick arteries were visible using energy subtraction. When the density of muscles was normalized to the minimum \( \# \text{black} \) utilizing I-K-edge CT, thick arteries were visible at high contrast.

Fig. 9 Tomograms of vials and a dog heart obtained using I-K-edge CT and a counter card. The image granularity was observed in both the tomograms without the FVC.
I media could be observed, the image granulations were observed. The image density analysis of the two glass vials with and without the FVC using the Image J is shown in Fig. 10. Compared with a tomogram with the FVC, it was difficult to observe image density difference between the two media without the FVC. Therefore, the count-rate smoothing is quite useful for improving the image quality under the low-count-rate imaging; the low-rate imaging is enabled to decrease the absorbed dose for patients and to increase the scan velocity without the pileups.

![Image density analysis of a glass vials filled with I media of two different densities. Compared with the tomogram without the FVC, the image density difference between the two media was large with the FVC.](image)

The projection data of a 30.0-mm-diameter lead rod positioned at the center on the turntable is shown in Fig. 11. We assume that the beams are parallel at 1.0 m from the X-ray source, and the X-ray transmissivity of the lead rod is approximated to be 0. In this case, the falling voltage \( V_f(t) \) and the rising voltage \( V_r(t) \) as functions of the time \( t \) from the FVC are written by

\[
V_f(t) = V_0 \exp\left(-t/\tau\right), \quad \left[ t = (\xi - 15)/25, \quad 15 \leq \xi < 45 \right]
\]

\[
V_r(t) = V_0 \left[1 - \exp\left(-t/\tau\right)\right], \quad \left[ t = (\xi - 45)/25, \quad 45 \leq \xi < 60 \right]
\]

where \( V_0 \) is the maximum output voltage from the FVC without X-ray absorption, \( \tau \) is the time constant of 50 ms, and \( \xi \) (mm) is the coordinate on the scan stage. Therefore, the two boundaries between the air and the lead rod can be observed clearly.

In our research, we developed a simple FVC consisting of a microcomputer, an integrator, and an operational amplifier to improve the image granularity in tomograms under low count rates, and the FVC was quite useful for smoothing the photon count rate. Although there are readily available FVC chips, an optimal electric circuit for the ED-CT should be selected to smooth the rate.

The image granularity in tomograms is primarily caused by the statistical photon-count error, and it is important to increase the photon number per measuring point to improve the image quality. Using this CT system with a scan velocity \( v \) of 25 mm/s, the photon count \( C \) per measuring point is written by

\[
C = R s/dv
\]

where \( R \) is the count rate and \( s \) is the scan step of 0.5 mm. In this CT system, the count rate for the I-K-edge CT was 1.9 kcps, and the maximum photon count per measuring point was calculated as approximately 40 counts. Therefore, the FVC with an integrator was quite effective to improve the image granularity. In addition, the two tomograms with two different energies for subtraction should be taken simultaneously to decrease the statistical errors in the subtraction image.

In this ED-CT system, the scan velocity was set to the maximum value of 25 mm/s to reduce the exposure time for CT. Lately, we have developed a high-speed X-ray scanner with a maximum velocity of 300 mm/s. Therefore, the exposure time would be reduced to approximately 1 minute when sufficient photon counts per measuring point for imaging can be obtained.

The pixel dimensions of the reconstructed CT image were 0.5 × 0.5 mm² because the scan step was 0.5 mm. However, the original spatial resolution is primarily determined by the pinhole diameter of 0.7 mm, and the spatial resolutions were 0.7 × 0.7 mm².

The image quality improves with decreases in the scan
and rotation steps. In addition, the time constants of the integrator in the FVC should be maximized to reduce image granulation primarily caused by the statistical photon-count error.

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 of Japan (MEXT). We also acknowledged Grant-in-Aid for Strategic Medical Science Research Center from MEXT, 2009-2013.

REFERENCES


