A Monochromatic X-Ray CT Using a CdTe Array Detector with Variable Spatial Resolution

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ABSTRACT

The CdTe semiconductor detector has a higher detection efficiency for x-rays and □amma rays and a wider energy band gap compared with Si and Ge semiconductor detectors. Therefore, the size of the detector element can be made small, and can be operated at room temperature. The interaction between a CdTe detector and incident x-rays is mainly photoelectric absorption in the photon energy range of up to 100 keV. In this energy range, Compton effects are almost negligible. We have developed a 256 channel CdTe array detector system for monochromatic x-ray CT using synchrotron radiation. The CdTe array detector system, the element size of which is 1.98 mm (h) x 1.98 mm (w) x 0.5 mm (t), was operated in photon counting mode.

In order to improve the spatial resolution, we tilted the CdTe array detector against the incident parallel monochromatic x-ray beam.

The experiments were performed at the BL20B2 experimental hutch in SPring-8. The energy of incident monochromatic x-rays was set at 55 keV. Phantom measurements were performed at the detector angle of 0, 30 and 45 degrees against the incident parallel monochromatic x-rays. The linear attenuation coefficients were calculated from the reconstructed CT images.

By increasing the detector angle, the spatial resolutions were improved. There was no significant difference between the linear attenuation coefficients which were corrected by the detector angle. It was found that this method was useful for improving the spatial resolution in a parallel monochromatic x-ray CT system.

Keywords: synchrotron radiation, variable spatial resolution, monochromatic x-ray CT

1. INTRODUCTION

Monochromatic x-ray CT has several advantages over conventional x-ray CT, which utilizes bremsstrahlung white x-rays from an x-ray tube. Images obtained from a monochromatic x-ray CT show two-dimensional distributions of linear attenuation coefficients. By using monochromatic x-rays with energies above and below contrast medium K absorption edge, it is possible to get an energy subtraction image. Quantitative CT images may be useful for the treatment planning in radiation therapy, which requires accurate knowledge of x-ray attenuation coefficients throughout the body, as well as for quantification of diagnosis.

We have developed a 256 channel CdTe array detector for a fan beam monochromatic x-ray CT system by using fluorescent x-rays generated by irradiating metal targets by synchrotron radiation[1-2]. In this CT system, the spatial resolution can be easily improved by changing the distance from the focal spot to the rotational center and to the detector, because the fluorescent x-rays are divergent. In the case of parallel-beam monochromatic x-ray generated by a monochromator, the spatial resolution is limited by the size of the CdTe detector element. Therefore, in order to improve the spatial resolution, we tilted the CdTe array detector against the incident parallel monochromatic x-ray beam.

2. MATERIALS AND METHODS

2.1. 256 channel CdTe Array Detector

The CdTe semiconductor detector has a higher detection efficiency for x-rays and □amma rays and a wider energy band gap compared with Si and Ge semiconductor detectors. Therefore, the size of the detector element can be made small, and can be operated at room temperature. The interaction between a CdTe semiconductor detector and incident x-rays is mainly photoelectric absorption in the photon energy range of up to 100 keV. Fig. 1 shows the relative contributions of photon interactions to the total attenuation coefficient for CdTe[3]. In this energy range, photoelectric effect is dominant and two energy peaks are measured. One is photoelectric absorption peak, and the other is K-escape peak. By measuring the photoelectric absorption peak, we can obtain an accurate intensity of an incident x-ray.

The 256 channel CdTe array detector has 256 CdTe semiconductor detector elements. The element size is 1.98 mm (h) x 1.98 mm (w) x 0.5 mm (t). Each element has an amplifier, two discriminators (an upper and a lower discriminator) and two 16-bits counters (an upper and a lower counter). By changing the discriminator level, we can get an incident x-ray energy spectrum. In a CT experiment, the upper and lower discriminator levels were set at upper and lower level of the photoelectric absorption peak respectively.

2.2. Phantom Experiments

The phantom experiments were performed at the BL20B2 experimental hutch in SPring-8. The distance between the light source and the experimental hutch is 210 m, and the divergence of the monochromatic beam is almost negligible in this experiment. The phantom consists of water, acrylic resin, Teflon, ethyl alcohol, acetone and paraffin. Fig. 2 shows a schematic diagram of the phantom. The energy of incident monochromatic x-rays was set at 55 keV. Phantom measurements were performed at the detector angle of 0, 30 and 45 degrees against the incident parallel monochromatic x-rays. By tilting the detector against the incident x-ray, the spatial resolution is improved by a magnification factor f, which is given by[4-5]

$$f = 1/\cos(\theta) \qquad (1)$$

where θ is the angle between the surface of the detector and the incident parallel monochromatic x-rays (Fig. 3).

A total of 360 projections covering 360 degrees were acquired, and CT images were reconstructed by using filtered back projection method. The linear attenuation coefficients were calculated from the reconstructed CT images.

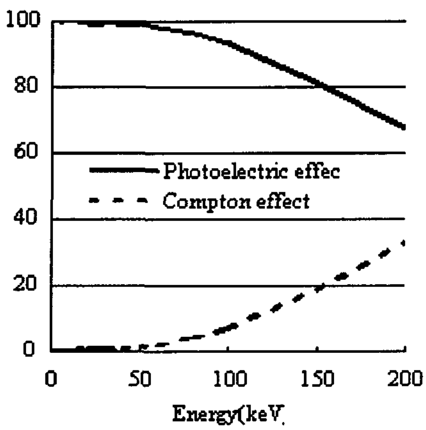


Fig. 1 Relative contributions of photon interactions to the total attenuation coefficient for CdTe

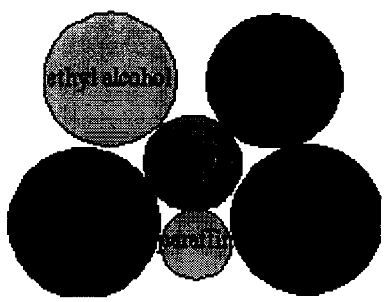


Fig. 2 Schematic diagram of the phantom

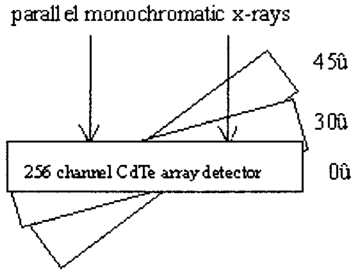
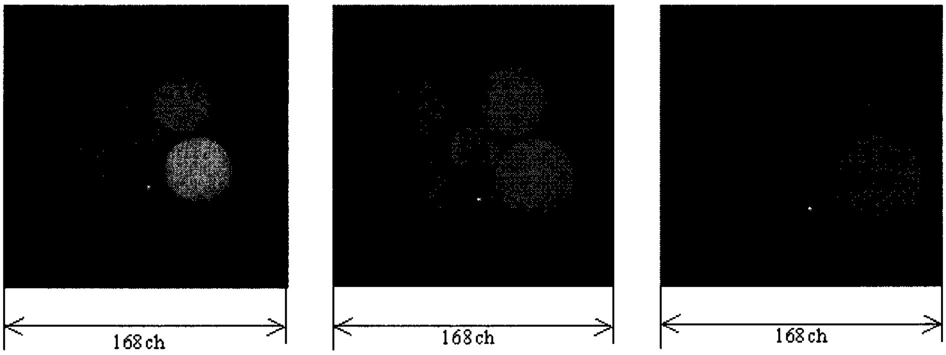


Fig. 3 Schematic diagram of the detector angle



(a) 0? (b) 30? (c) 40?

Fig. 4 Phantom images for different detector angles

3. RESULTS

Fig. 4 shows CT images of the phantom. By increasing the detector angle, the spatial resolutions were improved. The theoretical spatial resolution which was calculated form the size and the arrangement of the element is 2.5 lp/cm. The spatial resolutions for tilt angle of 30 and 45 degrees calculated form the phantom images were 3.0 lp/cm and 3.6 lp/cm, and the magnification factor f were 1.2 and 1.4 respectively. These values agreed with the theoretical values calculated from Eq. (1).

Fig. 5 shows linear attenuation coefficients which were corrected by the magnification factor as a function of detector angle. The linear attenuation coefficients were almost independent on the detector angle.

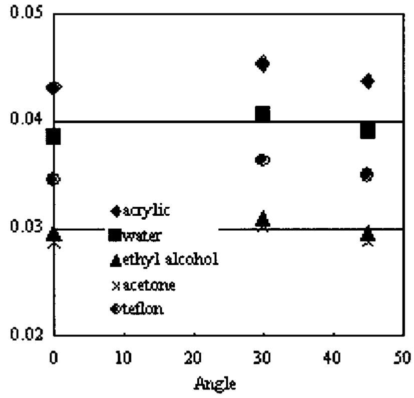


Fig. 5 Linear attenuation coefficients corrected for detector angles

4. DISCUSSION

In this experiments, a spatial resolution was improved by a factor of 1.4. By increasing the detector angle, it is expected that it will be further improved. However, the ratio of escape peak to photoelectric peak increases as the detector angle increases. This will cause the photon energy discrimination more and more difficult. Therefore, the improvement of the spatial resolution by a factor of two or three may be practical.

5. CONCLUSION

It has been shown that accurate linear attenuation coefficients can be obtained almost independent of the detector angle. It was found that this method was useful for improving the spatial resolution in a parallel beam monochromatic x-ray CT system.

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