Image quality evaluation of linear plastic scintillating fiber array detector for X-ray imaging

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Abstract It is important to assess image quality, in order to ensure that the imaging system is performing optimally and also identify the weak points in an imaging system. Three parameters mostly leading to image degradation are contrast, spatial resolution and noise. There is always a trade-off between spatial resolution and signal to noise ratio, but in scintillating fiber array detectors spatial resolution is not as important as signal to noise ratio, so we paid more attention to contrast and SNR of the system. By using GEANT4 Monte Carlo detector simulation toolkit, some effective parameters of the linear plastic scintillating fiber (PSF) array as an imaging detector were investigated. Finally we show that it is possible to use this kind of detector to take CT and DR (Digital Radiography) image under certain conditions.

Keywords Scintillating fiber, Detector quantum efficiency, Modulation transfer function, Image quality, GEANT4

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1 Introduction

Although plastic scintillator has fast response, more flexibility and strong electromagnetic immunity and has been used in a wide range of applications, such as particle tracking in high energy particle physics, calorimeters, etc., it has an problem for detection of X and gamma rays. The problem originates from the nature of this kind of organic scintillating material. On the other word, the material that is used as the core of the scintillating fibers has low Z and it is known that the interaction of X or gamma rays is strongly Z dependent. One of the most important interactions of X-ray with material is the Photoelectric Absorption (PA), of which probability is proportional to Z^5. X-ray in- cident upon a detector typically deposits only a fraction of its energy. The amounts of deposited energy are not always the same because of the stochastic nature of the X-ray interaction with matter. This phenomenon is the source of absorption noise in an imaging detector. As we know, the energy deposition of incident photon via PA can reach its maximal value, i.e., X-ray is fully absorbed in the detector. Therefore, less noise would be induced if the detector works in certain energy range of X-rays.

Capability of the plastic scintillating fiber (PSF) as an X-ray imaging detector has been proved and some important parameters of the detector have been extracted experimentally. For example the MTF curve for fiber array (with 50μm fiber diameter) was described.[1,2] In this work we attempt to find out some parameters of linear array of the PSF. Some of the parameters that are used in evaluation of image quality are DQE (Detector Quantum Efficiency), SNR (Signal to Noise Ratio) and MTF (Modulation Transfer Function). Any loss on these parameters at the detector level is irreversible and cannot be recovered by subsequent image processing. Radiation energy, dose level and also subject contrast as well as detector material and its size are the most effective factors determining the image quality. Linear array scintillating fiber detector with good collimating of the beam (as a fan shape) can be used for both CT scan and digital radiography. Advantages of this kind of detector, as

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compared with 2D array, are having less cross talk between fibers and less effect of scattering radiation. So, we would ignore these effects in our study.

2 Theory

Signal and noise properties for imaging systems can be described by detector quantum efficiency in frequency domain (DQE (*f*)). DQE (*f*) can be understood as the fraction of the number of quanta entering the imaging system effectively used by the system to produce an image. The DQE as a function of the spatial frequency of the object details is defined as

$$\mathrm{DQE}(f)=\frac{\mathrm{SNR}_{\mathrm{out}}^2}{\mathrm{SNR}_{\mathrm{in}}^2}=\frac{S^2\mathrm{MTF}^2(f)}{\Phi\mathrm{NPS}(f)} \quad (1)$$

where $\mathrm{SNR}_{\mathrm{out}}$ and $\mathrm{SNR}_{\mathrm{in}}$ are output (detected quanta) and input (incident quanta) signal to noise ratios, respectively, S is the average image signal, Φ is the incident X-ray flux, MTF is the modulation transfer function of the imaging system and NPS is the noise power spectrum of the image.

The MTF describes the signal attenuation as a function of the spatial frequency. In other words, spatial resolution of the detector is completely described by MTF that is defined as the modulus of the Fourier transform of the image of Line Spread Function (LSF).

$$\begin{aligned}\mathrm{MTF}(f)&=\left|\mathrm{FFT}\left[\mathrm{LSF}(x)\right]\right|\\&=\left|\frac{1}{\sqrt{2\pi}}\int_{-\infty}^{\infty}\mathrm{LSF}(x)\mathrm{e}^{2\pi ifx}\mathrm{d}x\right|\end{aligned} \quad (2)$$

3 Method

For the calculation of MTF, two methods could be used according to Eq.(2). One is to measure LSF image and take Fast Fourier Transform (FFT) of this image to get MTF. The other method for achieving MTF is via using Edge Spread Function (ESF) of the knife-edge imaging. By differentiating the ESF we can find LSF and then MTF would be obtained. Fortunately, for one dimension array we do not need to take care of high precision alignment of the target for these two methods of MTF calculation.

Although both techniques should give almost the same result, we define a phantom to check out both methods.

In order to calculate the noise power spectrum (NPS), we used uniformly exposed images to obtain r.m.s. fluctuations. Indeed the NPS (or Wiener spectrum) describes the variance in amplitude of each frequency component of a system.

4 Experiment setup

In this study we have used scintillating fiber, which emits green light with 492nm (i.e., 2.5eV) wavelength in peak of its light output spectrum. The used PSF is 0.25mm in diameter and 10cm in length. Scintillating core and cladding material are polystyrene (refractive index = 1.6) and acrylic (refractive index = 1.49), respectively. To irradiate the fibers we used several mono energy X (or gamma) rays. Source to Fiber Distance (SFD) is 20cm.

The phantom is made of tungsten and contains a slit with 250μm in width and 0.1mm in thickness. Fig.1 shows a schematic of the phantom and linear fiber array that was directly coupled to CCD (no any material is between them). CCD with 25μm pixel size has been chosen. Of course the purpose of using CCD in this simulation is only to count output light of the fiber and its distribution on CCD pixels, and finally the number of photons counted in each pixel would make matrix data for image reconstruction program. In addition it is supposed that the CCD's sensitivity to light photon is matched with output light spectrum of the fiber and also the CCD is front-illuminated type. The efficiency of the CCD has been ignored.

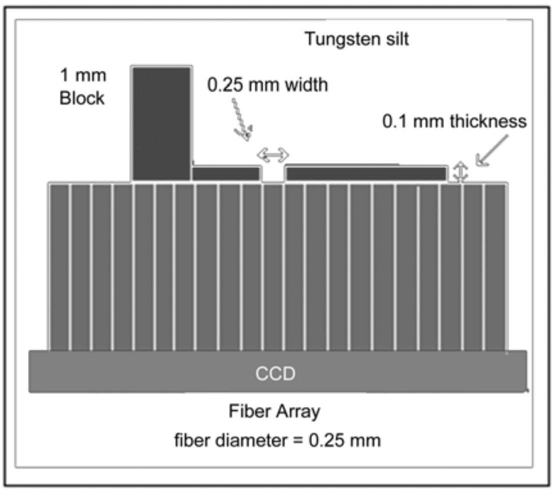

Fig.1 A phantom defined for simulation.

5 Results

The ability of an imaging system to transfer the latent image information is described by its detector efficiency (DE). DE also describes the contrast of the image somehow.

DE was calculated according to ratio of output light photon energy to incident photon energy.[3] Ignoring spatial frequency of the energy deposition within the fiber array, we have made the calculation of DE indeed for zero-frequency analysis. From the result shown in Fig.2, good contrast of the image in low energy of the incident beam is predicted. As is shown in this figure, DE has not reached to 75% even for low energy.

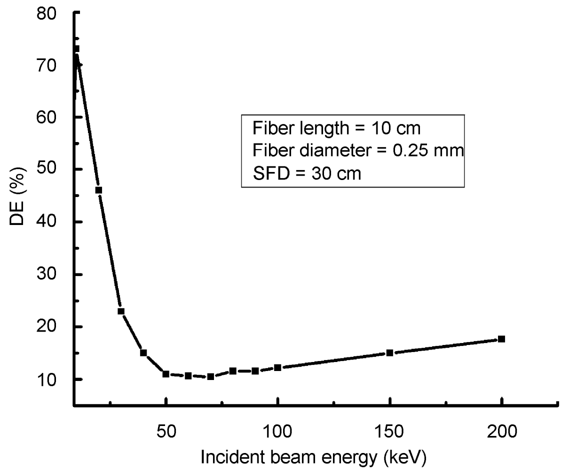

Fig.2 Detector efficiency as a function of incident beam energy.

As next step we irradiated our phantom by different energy of incident beam, the result is shown in Fig.3.

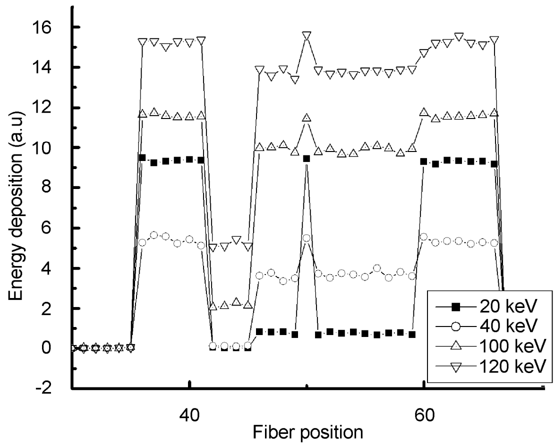

Fig.3 Image profile of the phantom for different incident beam energy.

The phantom is positioned at the entrance of the fiber array, hence the image magnification is 1.

The profile of the image shows that there is a better contrast in image for low subject contrast and the incident beam with 20keV. Fig.4 demonstrates the fact clearly. The image has been constructed by using output light of the fiber detected by CCD.

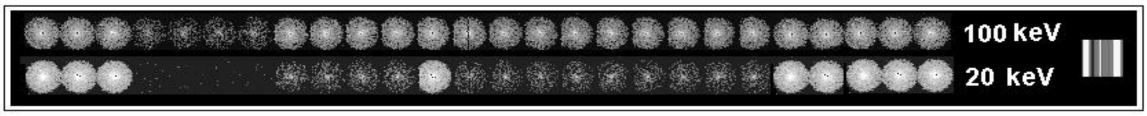

Fig.4 A profile of the image from the phantom created by linear fiber array.

The total length of the array shown in Fig.4 is only 6.5mm. The right corner of the image shows almost the whole size of the phantom as a radiograph in real size for 20keV of beam energy. The MTF calculated by ESF is shown in Fig.5.

Although Nyquist frequency of this imaging system is 2 lp/mm, the MTFs for different energies become very low for 0.5 lp/mm and above. However below this frequency MTFs show almost the same for four different energies except some small variation.

In order to calculate DQE(f), the Noise Power Spectrum (NPS) has to be known. The NPS was obtained by applying direct Fourier transform method along one dimension. The resulting plots are shown in Fig.6 for three different exposure levels.

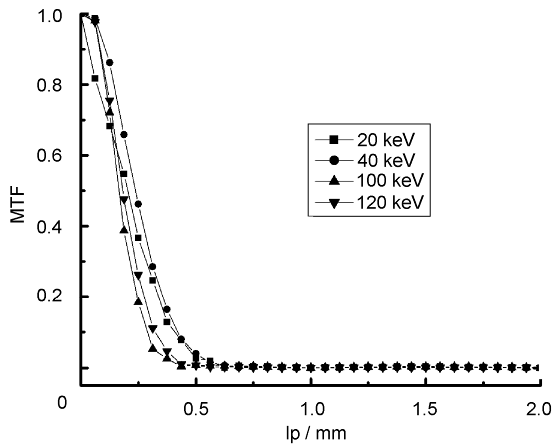

Fig.5 MTFs of the imaging system.

By using Eq.(1) the DQE(f) was calculated. The result is shown in Fig.7. Fig.7 shows that the DQE in spatial frequency for 20keV is better than that for 100keV, but both are going to become almost zero for frequencies above 0.6 lp/mm.

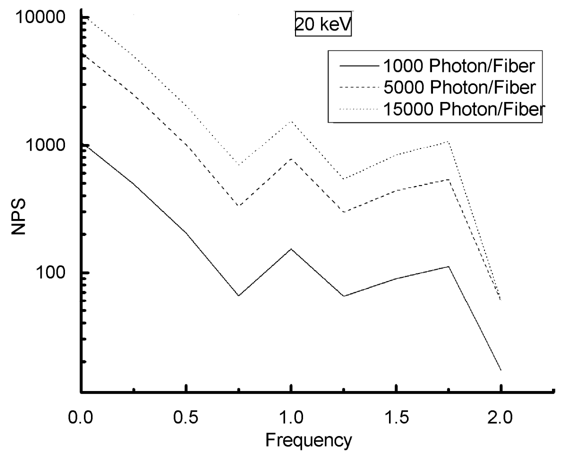

Fig.6 NPS at 20 keV for different exposure.

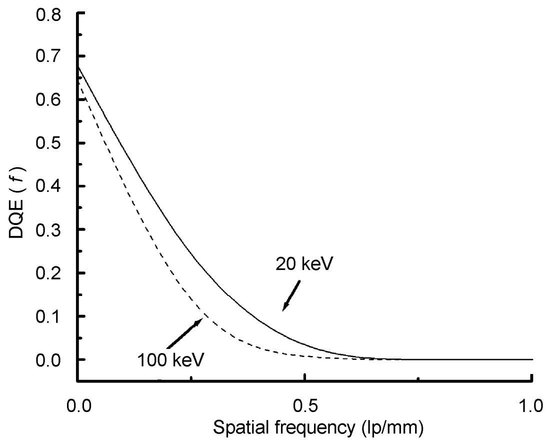

Fig.7 DQE (*f*) of the system for two different energies.

6 Conclusion

In this work, we demonstrated that the linear plastic scintillating fiber array has great potential as an imaging detector for 20-120keV energy range of electromagnetic radiation. From the measured DE, we find the best range of energy is below 40keV. Because of high attenuation coefficient of X (or gamma) ray for low energy, most of applications using this kind of imaging detector could be for things with even low subject contrast. The thickness of the detector or the fiber length (10cm) has no any significant effect to degrade or enhance the spatial resolution of the image in our energy range. The MTF for low energy X-ray seems to be a little bit higher than that for high energy X-ray. The results of MTF of the PSF array for different energies show that the spatial resolution is almost energy independent in the range mentioned above. For example, at 0.42 lp/mm the modulation transfer is 10% for 20keV while we have the same modulation transfer for 120keV at 0.3 lp/mm.

Due to trade-off between spatial resolution and detector efficiency, we must specially take care when the fiber is used for high energy, where the Compton scattering and unguided light (more light photons which increase crosstalk between fibers) have strong effect of degrading the spatial resolution of the image.

Finally, the real experiment results[2,3] show that the MTF of fiber bundle (2D array with fiber size of 50μm) is almost zero at 1 lp/mm and above. However, this has happened for the MTF of our detector (1D array) at almost 0.5 lp/mm but the size of our fiber is 5 times bigger. With respect to fibers size of two detectors, the resolution of our system should be five times poorer while it is only two times poorer than that reported for 2D array. This enhancement in resolution is due to the lack of cross talk when we use one dimensional fiber array. This could be an advantage of the system.

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