Characterization of X-ray Detector for CCD-based Electronic Portal Imaging Device


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(Received January 11, 2000. Accepted February 26, 2000)

Abstract: A combination of the metal plate/phosphor screen with a CCD camera is the most popular x-ray detector system among various electronic portal imaging devices (EPIDs). There is a need to optimize the thickness of the metal plate/phosphor screen with high detection efficiency and high spatial resolution for effective transferring of anatomical information. In this study, the thickness dependency on the detection efficiency and the spatial resolution of the metal plate/phosphor screen was investigated by calculation and measurement. It was found that the detection efficiency was mainly determined by the thickness of metal plate, while the spatial resolution was mainly determined by the thickness of phosphor screen. It was also revealed that the detection efficiency and the spatial resolution have trade-off in term of the thickness of the phosphor screen. As the phosphor thickness increases, the detection efficiency increases but the spatial resolution decreases. The curve illustrating the trade-off between the detection efficiency and the spatial resolution of the metal

INTRODUCTION

Cancer is one of the leading causes of death and a
plate/phosphor screen detector was obtained as a function of the phosphor thickness. Based on the calculations, a prototype CCD-based EPID was developed and then tested by acquiring phantom images for 6 MV x-ray beam.

Key words: Portal Image, EPID, Bremsstrahlung Spectrum, detection Efficiency, Spatial resolution, Frame averaging.

disease that reduces the quality of life in elderly popula-
tion. Radiation therapy is the most common method of controlling cancer among the several treatment options for cancer. The goal of radiation therapy is to deliver a prescribed dose as accurately as possible to a tumor region while minimizing the dose distribution to the neighboring normal tissues[1]. The standard course of treatment is divided into daily fraction of about 200 cGy dose, delivered over a period of 25–35 days for a total of 5000 to 7000 cGy cumulative dose[2]. Because of daily treatment, discrepancies in field placement occur frequently, such as patient movement, improper placement of shielding blocks, shifting of skin marks relative to internal anatomy and incorrect beam alignment[3]. As a result, recommendations by the International Commission on Radiation Units (ICRU) suggest that the accuracy in dose delivery should be ±5%[4]. For achieving this accuracy, the methods to reduce the frequency of discrepancies in field placement by frequent monitoring of patient positioning are developed, known as portal imaging. Portal imaging is a process to form an image of the patient during radiation treatment for ensuring that the correct region of the patient receives the radiation therapy and that the surrounding tissues are spared.

Currently, the most commonly used portal imaging method employs a radiographic film placed in contact with a metal plate. Unfortunately, these film images suffer from many disadvantages, such as low image quality, fixed display contrast, limited dynamic range and time delay due to film development[5]. The number of alternative detectors, known as electronic portal imaging device (EPID), have been developed, and each detector overcomes some of these limitations. The digital images not only open up possibilities for quantitative comparison with simulator images, but can also be processed to overcome poor contrast. And imaging can be performed in real-time. The video camera or CCD-based EPID is still the most popular detector system among various EPIDs. [5-8] (Fig.1) The x-ray detector consists of a metal plate and a phosphor screen. The metal plate generates high energy electrons when irradiated by therapeutic x-ray with energy in order of 4~25MV. And the phosphor screen converts electrons into visible light and the light diffuses through the screen and exits the rear surface of the x-ray detector which is viewed by video camera via a 45 mirror. The video signal is digitized and the digitized image can be viewed on a monitor. As a metal plate which acts a build-up region, brass, steel and copper with the range of 1~2.25mm in thickness are commercially used[8-9]. In the case of phosphor screen, a terbium doped gadolinium oxysulfide (Gd2O2S:Tb) screen which is commonly applied to diagnostic radiology, with the range of 150~500mg/cm² in coverage is used [8-9]. Metal plate/phosphor screen is largely responsible for the total performance of the video camera-based EPID because it is the first stage of transferring an anatomical information in the image chains of system. Therefore the optimal design of the metal plate/phosphor screen as an x-ray detector for portal imaging is needed. Two quantities, such as detection efficiency and spatial resolution, are considered as a touchstone of good performance. In this study the megavoltage x-ray spectrum from 6 MV linear accelerator (LINAC) and the characteristics of various combinations in thickness of the metal plate and the phosphor screen have been investigated by calculation and measurement. This result can be used to determine the optimal thickness of metal plate and phosphor screen for specific application. Finally, prototype CCD camera-based EPID was developed and images were acquired.

MATERIALS AND METHODS

1. X-ray Beam from Clinical LINAC

For the treatment of deep-seated tumors, megavoltage x-rays, typically in the range of 4 to 25MV, are required. LINAC is currently the most popular device for this application[10]. The energy spectrum of bremsstrahlung x-ray produced by the clinical LINAC is difficult to measure directly because detectors sensitive to photon energy cannot be used in the high photon flux produced by the accelerator. The flux is not adjustable and placing attenuators in the beam alters the spectrum. In order to obtain the bremsstrahlung spectrum from LINAC, Monte Carlo N-Particle, version 4B (MCNP4B) code[13] was used. The simulation geometry and dimensions of LINAC were designed based on the accurate geometrical data of 6MV LINAC (Siemens, Mevatron KD) to match the actual experimental setup. This consisted of an electron beam, a shielded and water-cooled gold target, a stainless
Fig. 1. Schematic diagram of video camera-based EPID

This steel flattening filter and ion chambers. For the incident electron beam, a 5-deg cone-shaped with kinetic energy of 5.58 MeV, incident perpendicularly on the target was considered. Electrons were followed to a cut-off kinetic energy of 0.5 MeV. This is sufficient, as the bremsstrahlung radiation yield of monoenergetic electrons in gold target with kinetic energies below 0.5 MeV is less than 5.3%. The number of emerging photons from 5,000,000 histories were collected into ten energy bins, each of width 0.558 MeV, if they landed in a circular region of radius 4.375 cm about the central axis at a distance 100 cm from the target. The energy spectrum was computed by multiplying the number of photons in a given bin by the energy at the center of the bin.

To confirm the simulation result, Schiff spectrum was also calculated using the following equation[11], this is a relatively simple analytical form and has been used extensively for estimating the spectrum shape from a high energy accelerator.

\[
\Gamma_{\text{mm}}(E, \nu) = 8 \left[ 2 \left( 1 - \frac{\nu}{E_0} \right) \left( \ln \frac{E}{E_0} - 1 \right) + \left( \frac{\nu}{E_0} \right)^2 \left( \ln \frac{E}{E_0} - \frac{1}{2} \right) \right]
\]

\[\text{Eq. 1.1}\]

\[
\epsilon = \left[ \frac{\mu \nu}{2E_n E} + \frac{\nu^2}{C} \right]^{1/2}
\]

\[\text{Eq. 1.2}\]

where

- \(E_0\): total energy of incident electron [MeV]
- \(E\): scattered electron total energy [MeV]

The Schiff spectrum presented here neglects several effects that contribute to the actual energy spectrum of LINAC, such as the electron energy spectrum within the target, Compton scattering of bremsstrahlung photons in the target or flattening filter, and scattering from collimator etc.

2. Detection Efficiency of X-ray Detector

In this study, the detection efficiency is defined as the total absorbed energy in the phosphor screen per incident x-ray. Since the number of optical photons generated is directly proportional to the total absorbed energy in a given phosphor screen. In order to determine the energy absorption within the phosphor screen, MCNP4B code [13] was used. In this study, the x-ray detector was simply modeled, which consisted of copper plate (0 ~ 50 mm) in contact with Gd:OxS layer (0.1 ~ 5 mm), with 20 cm in radius. Despite of complexity of phosphor screen material composition, the density of Gd:OxS, which is assumed to be homogeneous mono-layer, was reduced to 3.65 g/cm³ accounting for polyurethane polymer-binder and small air pockets within a realistic phosphor layer[9,14]. For the incident x-ray beam, a pencil beam with bremsstrahlung spectrum as calculated in the above section, incident perpendicularly on the x-ray detector was considered. Locally distributed energy absorption was estimated in cubic cells of phosphor layer in Monte Carlo simulation. The total absorbed energy was computed by adding the...
absorbed energy in a given cell.

To confirm the above calculation result, the system to measure the light output from metal plate/phosphor screen was developed. (Fig.2) It has a 35mg/cm²-coverage Gd₂O₂S (Lanex fast back screen, Eastman Kodak) attached onto a copper plate (0–50mm) as the x-ray detector. The detector is then optically coupled with PMT (Hamamatsu, PMT R5506) by an aluminum-coated, front-surface mirror with 45° inclined angle. The detector, mirror and PMT were placed in a light-tightened housing. The electronic signal from PMT is amplified by the current to voltage conversion amplifier (Hamamatsu, C1053-03) and then measured by both the pico-ammeter (Keithley, 485) and the oscilloscope. All experimental measurements were performed for the 6MV x-ray beam from LINAC (Siemens, Mevatron KD) at 300 MU/min and 100 cm SSD.

3. Spatial Resolution of X-ray Detector

To fully characterize the detector behavior it is necessary to have a model of optical photon transport out of the phosphor screen. The spatial resolution is defined by the full width at half maximum (FWHM) of a point spread function, which is the distribution of the optical photon- flux over the surface of the phosphor screen opposite to metal side. It is assumed that the optical photon generated in the center of a small cubic box of which total intensity is proportional to the total absorbed energy in the box and then propagates isotropically. Since the optical photon absorption-length (~4cm) in Gd₂O₂S[9] is sufficiently larger rather than the thicknesses of the generic Gd₂O₂S considered in this study, the attenuation of optical photons in the phosphor screen is neglected. In addition, the reflectance of the metal surface at the back of screen is also neglected because the light yield is not very sensitive to its value as described in [9]. The optical photon- flux collected over the solid angle of a rectangular pixel of size 2X×2Y can be expressed by [15–17],

$$P(z) = \int \int \frac{N_{opt}}{4\pi} \times \frac{z}{(x^2 + y^2 + z^2)^{3/2}} dxdy$$  \hspace{1cm} (Eq.3.1)

where x, y and z are 2-dimensional coordinates of the end-sided phosphor screen and distance to the light source, respectively. \(N_{opt}\) is the number of photons generated at which the x-ray energy is absorbed within phosphor screen and can be calculated by

$$N_{opt} = \varepsilon \times \frac{E_{opt}}{E_{rad}}$$  \hspace{1cm} (Eq.3.2)

In Eq. 3.2, \(\varepsilon\) is the intrinsic conversion efficiency in the range of 15–20% [18–19], \(E_{opt}\) and \(E_{rad}\) are the optical photon energy (2.28keV for 545nm wavelength from Gd₂O₂S:Tb) and the locally absorbed energy within phosphor screen from the secondary fast electrons, respectively.

Locally distributed energy absorption was estimated in 100×100×100μm³ cubic cell of phosphor layer by Monte Carlo simulation as described in the previous section. Then, the center point of the cell was assumed to be the representative position as the optical photon-source point. Based on the Monte Carlo simulated spatial distribution of energy absorption, the distribution of optical photon flux collected by 10×10μm² pixel area as a function of pixel position on the free surface of the phosphor screen was calculated by Eq. 3.1 and Eq. 3.2 using a hand-made C program. To confirm the calculated spatial resolution, the experimental measurement was carried out. The line spread function is equal to the integral of the point spread function and the derivative of the edge spread function[20]. To determine the edge spread function of metal plate/phosphor screen, a density profile of a film across an edge was measured. A lead block with dimensions of 20×10×5cm³ was placed on a film cassette and across half of the x-ray field of 6 MV LINAC (Siemens, Mevatron KD) with 167cm SSD. The 125mg/cm²-coverage Lanex screen attached onto a 2mm-thick copper plate was placed in the film cassette with the phosphor layer of the detector in contact with the portal film (Kodak, X-Onlat V film) which is commonly used for the therapy verification. The film was exposed by the x-ray intensity of 50 MU (Monitor Unit, 1MU may be calibrated to equal 1 cGy at 100cm SSD for a 10×10μm² field size at 5cm-depth in water). Therefore, the film recorded the edge spread function of the metal plate/phosphor screen detector while negligibly influencing the spread. The shape of the edge spread functions were measured with film digitizer passing through the image of the edge, perpendicular to the edge length. Thirty scans were made and these scans were averaged together to reduce uncertainty in the final estimation of the edge spread function.

4. Prototype CCD-based EPID

In order to capture the therapeutic x-ray images, a
prototype CCD-based EPID that consists of the metal plate/phosphor screen detector and CCD camera have been developed based on the calculation results. As an x-ray detector, the Lanex screen with 135mg/cm² coverage bonded onto a 2mm-thick copper plate is used. The detector is viewed by a CCD camera with a field of view (FOV) of 20×20cm² using a 45 inclined aluminum-coated, first-surface mirror. Samsung B2-230EID operated by the interlaced scanning mode was used as a CCD camera with the 6-12mm zooming c-mo to zoom lens. These all the components were placed in the light tightened housing. The PCI-1408 Imaq board (National Instrument) as a frame grabber was used for data acquisition. For the 6 MV x-ray beam from LINAC (Siemens, Mevatron KD) at 300MU/min and 100cm SSD, all measurements were performed with the humanoid head phantom. (Fig. 3)

**RESULTS AND DISCUSSIONS**

1. Bremsstrahlung Spectrum

Simulated bremsstrahlung spectra, number and intensity and Schiff intensity spectrum were represented. (Fig.4) The mean energy of the bremsstrahlung x-ray is 1.57 MeV. As shown in Fig.4, the Monte Carlo simulation result provides reasonable agreement with a theoretical model and may be useful for applications in which the exact spectrum is not critical. The normalized number spectrum is, then, implemented to the source input of Monte Carlo simulation for estimating the detection efficiency of metal plate/phosphor screen detector.

2. Detection Efficiency

For the incident bremsstrahlung x-ray with the mean energy of 1.57MeV, the detection efficiency is shown in

![Graph](image)

**Fig. 3.** Experimental setting of prototype CCD-based EPID for measurement of humanoid head phantom images in the 6MV LINAC

**Fig. 4.** Simulated bremsstrahlung spectra, number (dot histogram) and intensity (line histogram) and Schiff intensity (line) spectrum were represented

**Fig. 5.** Detection efficiency for 0.3, 1 and 3 mm-thick phosphor as a function of the metal thickness

**Fig. 6.** Relative electron flux emitted from the metal plate as a function of metal thickness

Fig. 5 in unit of percentage for the 0.3, 1, 3mm-thick phosphor screen as a function of the metal thickness. The result shows that, the detection efficiency rapidly approaches to the maximum value at a range of 1.5~2.0mm of metal thickness and then decreases very slowly as the metal thickness increases. This characteristic can be analyzed by using the electron flux shape emitted from the metal plate (Fig.6) and the electron path length in the phosphor screen. The number of electrons penetrating metal plate depends upon their energies and corresponding path lengths, so that the total absorbed energy distribution in the phosphor screen as a function of metal thickness shows a maximum value. The calculated detection efficiency was compared using experimental measurement result. Fig.7 shows the normalized light yields of calculation and measurement as a function of the metal thickness. The presence of the copper plate enhances the absorbed energy of the phosphor screen, particularly for the thinner phosphors. The phosphor screen is close to the electron equilibrium throughout its thickness if the metal is sufficiently thick. In other words, the metal plate increases the energy absorption of the 0.3mm-thick phosphor screen by about 10 times while for 3mm-thick phosphor screen the enhancement is only about 1.2 times. From this result, it was found that, the improvement of detection efficiency due to the metal plate is more effective, as the phosphor thickness is thinner.

3. Spatial Resolution

The distribution of the optical photon flux collected by 10×10m² pixel area on the free surface of the phosphor screen as function of the pixel position was calculated. Fig.8 shows the results for various phosphor thickness without metal plate, and Fig.9 represents the results for 1 mm-thick phosphor layer with various thickness of the metal plate. As shown in figures, it is observed that the spatial broadening is largely affected by the phosphor thickness but the metal plate could be negligible, in other words, the metal plate does not degrade the spatial resolution. From the results, it can be inferred that electrons emitted from metal plate is not significantly spread, but is well confined along the direction line of the incident X-ray. Supplementary Monte Carlo simulation showed that 90% of electron flux emitted from the metal plates of thickness from 1 to 5mm is confined within the area of 1mm-diameter.

To quantify the spatial resolution, the edge spread function was fitted with the second-order exponentially decay curve as obtained from portal film. For comparing the measured edge spread function and the calculated
point spread function, these are converted into line spread functions by differentiation and integration, respectively. These two line spread functions are shown in Fig.10, and are agreed well to each other in error boundaries. The FWHMs of the line spread functions appear to be 0.342 mm and 0.436mm, for calculation and measurement, respectively. Unlike the calculation model, the phosphor screen is supported on a plastic support[21] and is bonded to the metal plate by plastic tapes so that the phosphor is not in direct contact with the metal plate in measurement. And also the scattering effects of the incident x-ray by surrounding materials including the film cassette were not considered in calculation. Since the high energy electrons emitted from the metal plate can spread in the plastic support and tapes before entering the phosphor and the film can be exposed by the scattered radiation, so that the measured FWHM is larger than the calculated one.

4. Acquired Images and Frame Averaging Method

In this experiment, 60 frames of images were acquired in video speed of 30 frames per second. As shown in Fig.11(a), an acquired single-frame image suffers notorious quantum noise, which is due to the counting statistics of the incident x-rays, the conversion fluctuations of x-rays into light quanta in x-ray detector and signal formation fluctuations in CCD camera[22]. The quantum noise can be largely reduced by the frame averaging method and Fig.11(b) shows the image which was acquired after 60 frames averaging. As the number of averaged frames increases, SNR (Signal-to-Noise Ratio) increases. At 60 frames averaged image, SNRs are increased by 10 and 17 times for the skull and neck region, respectively. In addition, between the sampled areas, the image contrast rapidly increases as the number of frames averaged increases, and then starts to saturate at around 8 frames. (Fig. 12)

CONCLUSION

The detection efficiency as well as the spatial resolution for various combinations of metal plate/phosphor screen as a detector of video camera-based EPID were estimated by calculations and verified by the experimental measurements. The results show that, the detection efficiency is maximized by use of the 1.5–2.0mm-thick metal plate, and is slowly decreased as the metal thickness increases over 2mm. And it was found that the detection efficiency strongly depends on the emitting electron flux from the metal plate. The spatial resolution is degraded

as the phosphor thickness increases, but the degradation of the spatial resolution due to the metal plate can be negligible because of the strong directionality of electrons. It was revealed that the detection efficiency and the spatial resolution were mainly determined by the thickness of metal plate and phosphor screen, respectively.

Since the optimal thickness of the metal plate is 1.5~2.0 mm, the detection efficiency and the spatial resolution have compromise in terms of the thickness of the phosphor screen. As the phosphor thickness increases, the detection efficiency increases but the spatial resolution decreases. The curve illustrating the trade-off between the detection efficiency and the spatial resolution of the metal plate/phosphor screen was obtained as a function of the thickness of phosphor screen with the 2mm-thick copper plate. (Fig.13) The spatial resolution of the portal film is 0.87 mm, and the lowest detection efficiency of commercially used x-ray detector of EPID is 0.24%. This curve can be used to determine the optimal thickness of metal plate/phosphor screen as the x-ray detector of electronic portal imaging system as well as another imaging system using the similar detector structure.

The prototype CCD-based EPID with FOV of 20x20 cm² was developed and therapeutic x-ray radiographs were acquired. Although, in the interlaced mode, the captured image suffered from serious quantum noise, the quantum noise was reduced by frame averaging method. The frame averaging method improved the SNR and the image contrast. In order to reduce the quantum noise and to improve the image quality in therapeutic x-ray imaging application, the frame averaging is necessarily required.

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