Simulating the Emission Efficiency and Resolution Properties of Fluorescent Screens By Monte Carlo Methods

Panagiotis F. Liaparinos, Ioannis S. Kandarakis, Dionisis A. Cavouras, Dimitris N. Nikolopoulos, Ioannis G. Valais and George S. Panayiotakis

***Abstract*—The aim of this study was to predict Quantum Detection Efficiency (QDE) and Modulation Transfer Function (MTF) of granular phosphor screens, used in detectors of medical imaging systems, using Monte Carlo methods. The transfer of energy through the screen was modeled as a series of energy converting stages, assuming that the screen was divided into a number of elementary thin layers of specific thickness. Screen characteristics were predicted considering either polychromatic or monochromatic x-ray beam in the energy range used in mammographic applications. The screen coating weight was selected to be 31.7 mg/cm^2, which corresponds to a screen thickness of 85 μm. X-ray photon interaction site was determined by calculating the probability mass function for each layer, separately. Light quanta were considered to be isotropicaly created within phosphor's mass. The number of light quanta generated per absorbed x-ray photon was calculated using the intrinsic x-ray to light conversion efficiency. Light absorption and light scattering were assumed to be the possible mechanisms for light attenuation. Results were verified using previous data for Gd_2O_2S:Tb phosphor material. QDE and MTF were found in close agreement to analytically calculated data published in two previous studies. The agreement between Monte Carlo and theoretical values was better than 0.2% and 5% respectively.**

P. F. Liaparinos is with the Medical Physics Department, University of Patras, 26500 GREECE (e-mail: liapkin@hotmail.com).

I. S. Kandarakis is with the Department Medical Instruments Technology of Athens, 12210 GREECE (e-mail: kandarakis@lteiath.gr).

D. A. Cavouras is with the Department Medical Instruments Technology of Athens, 12210 GREECE (e-mail: cavouras@lteiath.gr).

D. N. Nikolopoulos is with Department the Medical Instruments Technology of Athens, 12210 GREECE (e-mail: dnikolop@med.uoa.gr).

I. G. Valais is with the Medical Instruments Technology of Athens, 12210 GREECE (e-mail: valais@teiath.gr).

G. S. Panayiotakis is with the Medical Physics Department, University of Patras, 26500 GREECE (e-mail: panayiot@upatras.gr).

I. INTRODUCTION

PHOSPHOR materials are employed as radiation to light converters in detectors of x-ray or gamma ray medical imaging systems [1], [2]. Phosphors are often used in the form of screens, which are coupled with optical detectors (films, photocathodes etc) [3]. In most x-ray imaging applications phosphor screens consist of a large number of phosphor particles (grains) embedded in a binding material (granular screens). In designing medical imaging detectors, the x-ray absorption and image resolution properties of phosphor screens are of crucial importance. X-ray absorption and spatial resolution, are often expressed by the quantum detection efficiency (QDE) and the modulation transfer function (MTF), respectively [4]. These two parameters are associated with the intrinsic x-ray and optical properties of phosphor materials [4], [5]. Since, in some cases, these properties cannot be accurately estimated by experimental techniques, Monte Carlo methods may be applied [6]. Monte Carlo methods, which conveniently describe the random nature of physical phenomena, can be successfully applied to simulate radiation transport within the phosphor material. Monte Carlo techniques have been previously applied in x-ray imaging to investigate either detector performance in general radiography or patient dose estimation: Morlotti investigated the emission efficiency and resolution properties of phosphor screens under conditions used in general radiography [7]. Chan and Doi studied the effects of scattered radiation in phantom doses and in image quality [8]. Boone et al performed simulation for the effect of K-fluorescence in various detector materials [9]. Carlsson et al have used Monte Carlo techniques to estimate energy imparted to the patient and image receptors [10].

In the present study, a new Monte Carlo code using MATLAB platform was developed in order to predict QDE and MTF of phosphor screens at x-ray energies used in mammography. Simulation was performed by taking into account the x-ray absorption, light production and light propagation properties of phosphor materials and by considering that the phosphor screen was divided into a series of elementary x-ray absorbing thin layers. Results were

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validated by comparison with published data concerning a commercially available phosphor screen [5].

II. Materials and Methods

A Monte Carlo program was developed to investigate the x-ray absorption and image quality properties of phosphor screens. The simulation of x-ray and light interactions was based on suitably sampling cumulative probability distribution functions (CDF) using random numbers obtained from random number generators [6]. Simulation was suitably modeled in order to allow for variability of the following components: phosphor screen thickness and dimensions, material composition, x-ray energy spectrum and the number of photon histories within predetermined conditions. The method was applied to examine the properties of the commercially available Min-R Kodak screen consisting of Gd_2O_2S:Tb phosphor

A. Geometry of the model

The geometry of the model is illustrated in figure 1. The phosphor screen was divided into a number N of superimposed thin grain layers, i.e. thin layers of thickness equal to that of the mean size of grains $(7\,\mu m)$. A pencil x-ray beam was considered to impinge on the center of screen surface. An x-ray photon interacts within a thin phosphor layer Δx at depth $x = i\Delta x$, $i = 1,\ldots N$ and light photons were generated isotropicaly towards all directions. The coating weight of the screen was selected to be $x_0 = 31.7\,mg/cm^2$ of Gd_2O_2S phosphor corresponding to a thickness of $85\,\mu m$, which is equal to that of commercially available Min-R Kodak screen [5]. In addition a phosphor packing density of 50% was assumed.

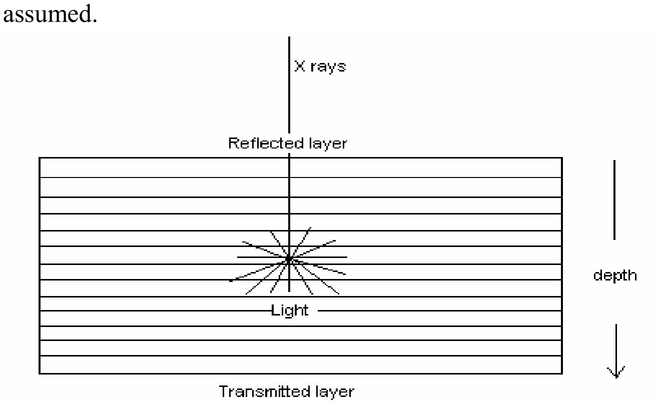

Fig. 1. Diagram illustrating the geometry of the model.

B. Input data

X-ray interaction with the phosphor material was expressed through the exponential attenuation law governed by the corresponding x-ray attenuation coefficients. Numerical values of the x-ray attenuation coefficients for the chemical elements of the Gd_2O_2S:Tb phosphor material were obtained from tabulated data of Storm and Israel [11]. The total mass attenuation coefficient of the material, at x-ray energy E, was calculated by the following equation:

$$\frac{\mu}{\rho}(Z_{mat}, E) = \sum_i w(Z_i)\frac{\mu}{\rho}(Z_i, E) \qquad (1)$$

where $w(Z_i)$ is the fractional weight and $\frac{\mu}{\rho}(Z_i, \rho)$ is the mass attenuation coefficient of the i element in the phosphor material, respectively [12]. Intermediate values of the mass attenuation coefficients corresponding to all energy values of the x-ray spectrum were obtained by the interpolation method.

A typical mammographic x-ray spectrum (molybdenum anode-molybdenum filter) was used. Data were taken from tabulated values corresponding to a molybdenum anode x-ray spectrum model [13]. The x-ray spectrum was generated by sampling the distribution arising from tabulated data. It was assumed that the random variable E, corresponding to the x-ray photon energy, can take discrete values E_j, where $j = 1,2,3,\ldots N$ with point probabilities $p_1,\ldots, p_N$, respectively. A corresponding cumulative distribution function was used which was expressed as follows:

$$P(E_j) = \sum_{j=1}^{j} p_j \qquad (2)$$

The sampling of the x-ray spectrum is given by the following formula:

$$E = E_j \quad \text{if} \quad P(E_{j-1}) \le R_1 \le P(E_j)$$

where R_1 is a random number uniformly distributed in the interval (0, 1].

C. Determination of interaction site and type of x-ray photons

A history starts when an x-ray photon, with energy obtained from the energy distribution of x-ray spectrum, impinges vertically on the center of the phosphor screen surface. Due to the high effective atomic number of the phosphor material and the relatively low energy of x-ray photons, x-ray scatter was considered to be of negligible contribution. Hence photoelectric effect was assumed to be the principal x-ray interaction. An x-ray absorption event was determined using a function corresponding to the quantum detection efficiency of the phosphor and a random number R_2, uniformly distributed in the interval (0, 1], as follows:

$$QDE(E, x) = 1 - \exp\left(-\frac{\mu(\mathrm{E})}{\rho}x\right) \qquad (3)$$

If $R_2 \le QDE(E, x)$ then the x-ray photon is absorbed.

where $QDE(E, x)$ is the quantum detection efficiency of the phosphor screen, $\frac{\mu(\mathrm{E})}{\rho}$ is the mass attenuation coefficient of

the medium at the energy E of x-ray photon, ρ is the density of the material and x is phosphor screen thickness.
The probability function:

$$PF_i(E) = \exp(-\frac{\mu(\mathrm{E})}{\rho}(i-1)\rho_{\mathrm{P}}\Delta x) - \exp(-\frac{\mu(\mathrm{E})}{\rho} i\rho_{\mathrm{P}}\Delta x) \qquad (4)$$

was calculated for each layer separately to indicate the layer where the x-ray photon was absorbed [14]. $PF_i(E)$ is the absorption probability of the x-ray photon, ρ is the bulk density of the phosphor, ρ_p is the packing density and Δx is the layer thickness. The probability mass function was determined by normalizing the probability function:

$$\mathrm{Pr}_i = \frac{PF_i}{\sum_{i=1}^{N} PF_i} \qquad (5)$$

Generating a random number R_3 uniformly distributed in the interval (0, 1]. If

$$\sum_{i=1}^{i-1} \mathrm{Pr}_i \le R_3 \le \sum_{i=1}^{i} \mathrm{Pr}_i$$

the x-ray photon was absorbed in i layer.

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D. Light production

The number of light quanta produced per absorbed x-ray of energy E was determined using the following relationship [5]:

$$G(E) = \frac{En_c}{\mathrm{E}_\lambda} \qquad (6)$$

where E is the energy of absorbed x-ray photons, n_c is the x-ray to light conversion efficiency of the phosphor screen and E_λ corresponds to the mean energy of the emitted light quanta for the material under investigation, which was assumed to be equal to $2.4 \times 10^{-3}\,keV$. The emitted light photons were radiated isotropically.

2.5. Determination of free path length of light photons

When a light photon was emitted, its free path length x_l was obtained from the following exponential probability density function [7]:

$$pdf\ (x_l) = \frac{\mu}{\rho}\exp(-(\frac{\mu}{\rho})\rho_P x_l) \qquad (7)$$

by a random sampling using an inversion method [8] such that:

$$xl = -(1/(\frac{\mu}{\rho})\rho_P)\ln R_4 \qquad (8)$$

where μ is the linear attenuation coefficient of light photon and R_4 is a random number uniformly distributed in the interval (0, 1].

E. Light Propagation

Light scattering on phosphor grains and light absorption within the phosphor mass were assumed to be the main interactions of light photons during light propagation through the screen material. Light interactions have been previously described by specific optical coefficients σ and β , which were directly related to the absorption (a) and scattering (s) coefficients of optical photons by the following relations [3]

$$\sigma = [a(a+2s)]^{1/2} \quad \text{and} \quad \beta = [a/(a+2s)]^{1/2} \qquad (9)$$

The values of the optical coefficients were taken to be equal to $\sigma = 11\,mm^{-1}$ and $\beta = 0.22$ according to a previous study [5]. Coefficients a and s were found equal to $2.42\,mm^{-1}$ and $47.58\,mm^{-1}$ respectively, using (9). The relative probability of light photon absorption may be written as follows:

$$P_a = a/(a+s) \qquad (10)$$

A random number R_5 is generated and if $R_5 \le P_a$ light absorption takes place otherwise light scattering will occur. It was assumed that the scattering angle follows an isotropic distribution.

F. Output

The history of x-ray or light photons was terminated when all of their energy was absorbed within the phosphor screen or escaped from the material structure. Point spread function (PSF) and modulation transfer function (MTF) were calculated from the number of light quanta at each position of the screen output. PSF was determined by the spatial position of light quanta and MTF curve was calculated using the Discrete Fourier Transform (DFT) of PSF.

III. Results and Discussion

Monte Carlo calculations have been performed for a large number of x-ray photons. The validity of our simulation program was first verified by the accurate description of the x-ray absorption within the phosphor screen. Figure 2 shows the variation of the absorbed fraction of x-ray photons with respect to screen thickness. 10^6 histories from monochromatic x-ray photons of energy $20\,keV$ were tracked. Points represent sampled values obtained by Monte Carlo simulation between two successive layers while solid line represents data corresponding to the theoretical relative probability of x-ray absorption to each layer, which is given by (5). The two distributions were found very similar and the absolute value of the average relative difference between Monte Carlo (MC) and theoretical values (TH) was estimated to be 0.38%.

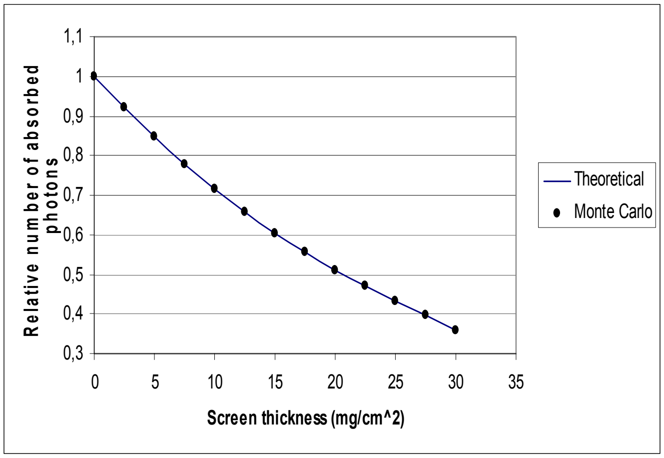

Fig. 2. Theoretical and Monte Carlo variation of the relative number of absorbed x-ray photons with respect to screen thickness.

Figure 3 shows the comparison between the x-ray spectrum incident on the surface of the screen and the absorbed x-ray spectrum. Each spectrum was normalized to represent a probability density function. The ratio of absorbed photons per incident photons or the quantum detection efficiency of phosphor screen was found equal to:

$$\frac{Absorbed \quad photons}{Incident \quad photons} = \frac{831280}{1000000} = 83.13\,\%$$

This value was compared with the theoretical value obtained by the following analytical model:

$$QDE\ (E,x) = \frac{\int_{E_1}^{E_2} [1 - exp(-\frac{\mu(E)}{\rho} x)] S_X(E)dE}{\int_{E_1}^{E_2} S_X(E)dE} \qquad (11)$$

where E_1 and E_2 are the lower and upper energy limits of the x-ray spectrum and $S_X(E)$ is the number of x-ray photons per energy interval dE. The theoretical value was found equal to 83.20%. The relative difference (MC-TH)/TH between Monte Carlo (MC) and theoretical value (TH) was 0.2%, which is may be considered negligible.

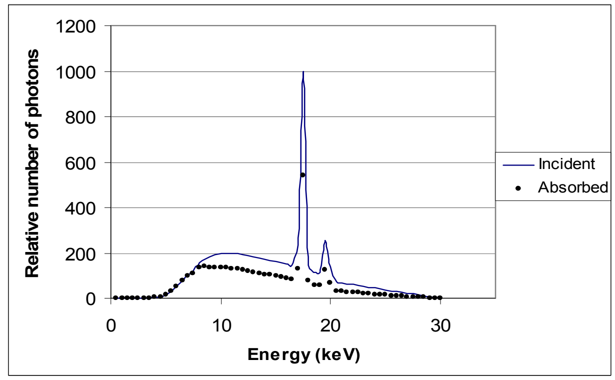

Fig. 3. Comparison between incident and absorbed x-ray spectrum.

Figure 4 illustrates the MTF curve of Gd_2O_2S:Tb screen as a function to the spatial frequency. The curve of our study was compared with previously published curves obtained by fitting theoretical model equations to experimental data [5]. The curves correspond to $31.7\,mg/cm^2$ coating weight using a $30\,kVp$ poly-energetic molybdenum x-ray spectrum, filtered by $0.0051\,mm$ of molybdenum and $4.2\,cm$ of Lucite to simulate beam hardening by an average breast [5]. All curves correspond to transmission mode of observation i.e. light emitted from the rear non-irradiated screen side was considered . This mode simulates digital radiographic detector configuration. The agreement was better than 3.5% when compared to data obtained by Nishikawa and Yaffe [5] and 5% when compared to data obtained by Kandarakis et al [3].

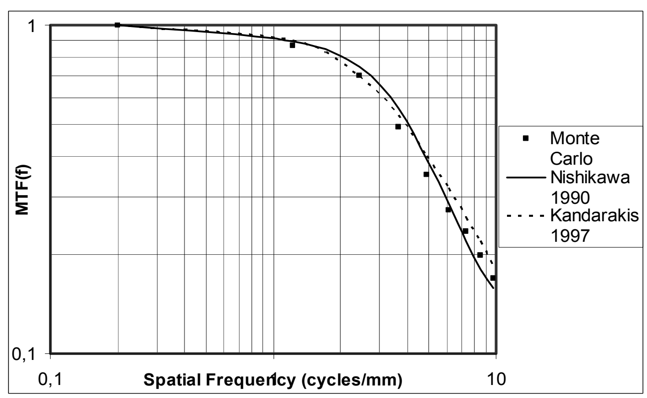

Fig. 4. Comparison of MTF(f) as predicted by Monte Carlo studyand as calculated by theoretical models.

Figure 5 shows the comparison of MTF for x-ray photon energies of $20\,keV$, $30\,keV$ and $40\,keV$ respectively. MTF becomes higher with increasing x-ray energy. X-ray scattering was not taken into account since for materials and x-ray photon energies considered in our study, it was assumed to be negligible. This was physically expected since higher energy x-ray photons penetrate deeper within the phosphor's mass. Light photons are thus generated closer to the screen's output surface. This decreases light spread and improves spatial resolution.

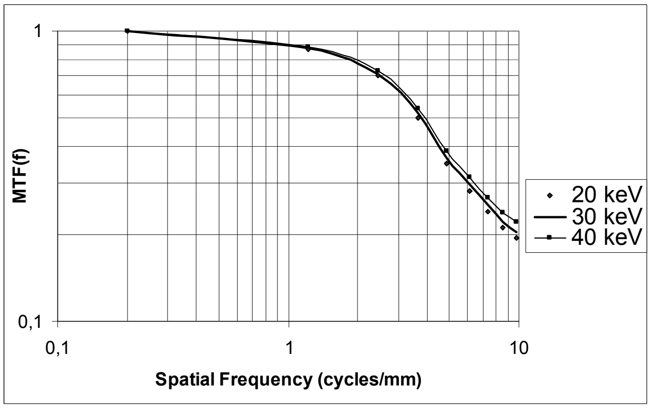

Fig. 5. Comparison of MTF (f) for energies 20 keV, 30 keV and 40 keV.

IV. CONCLUSIONS

In the present study a Monte Carlo program, using MATLAB platform, was developed to provide the distribution of light quanta and to predict the quantum detection efficiency and the modulation transfer function of x-ray phosphor screens. The present Monte Carlo study was based on a simple physical model describing the x-ray and light photon interactions within the phosphor material, giving low statistical error even if a relatively small number of x-ray histories is applied The verification of the present Monte Carlo study was carried out by comparison between results obtained from Monte Carlo calculations and previous theoretical calculations for $Gd_2O_2S:Tb$ phosphor screen under similar conditions. Results were found to be very similar confirming the validity of our Monte Carlo simulation.

V. ACKNOWLEDGMENT

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VI. REFERENCES

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