Scintillators in X-Ray Imaging:The Miscirlu Project

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Abstract

Luminescent materials are used as X-ray detectors of medical imaging systems. Out of a large variety of materials, Terbium (Tb)-activated phosphors and needle-like columnar structured CsI:Tl phosphors are currently the most widely used ones. Parameters commonly used to assess the imaging performance of luminescent materials are: the Quantum Detection Efficiency, the Luminescence Efficiency (LE), the Optical Spectral Distribution (OSD), the Modulation Transfer Function (MTF), the Noise Power Spectrum (NPS), the Detective Quantum Efficiency (DQE) and the Information Capacity (IC). The scope of the MISCIRLU project is an in depth theoretical and experimental analysis of the performance of scintillator materials as X-ray detectors. The theoretical analysis is performed through Mie scattering theory, Monte Carlo simulation and analytical modeling. The experimental analysis is performed through LE, OSD, MTF and NPS measurements. The initial results of the MISCIRLU project are focused in the theoretical analysis of the luminescent materials like the effect of the grain size, detector thickness, activator importance and scintillator crystal intrinsic conversion efficiency non-proportionality. In addition MTF and NPS have been evaluated via free software tools.

Key words: Scintillators, Medical imaging, Detectors.

Introduction

Diagnostic medical imaging systems are based on recording and storing information on both the anatomy and pathology of the human body. Specifically, ionizing radiation interacts in the body of the patient and then detected using appropriate X-ray detectors. These detectors have the ability to absorb ionizing radiation and to convert it to secondary information carriers. The carriers may be electrons in direct detection detectors or optical photons in detectors containing scintillator materials also known as phosphors (indirect detection). The phosphors have the ability to convert ionizing radiation into light photons (i.e. X-ray luminescence). The light emitted is incident on the surface of an optical detector, which is sensitive to the optical photon energy spectrum. These optical information carriers contribute to the formation of medical images. The final image is characterized by parameters which reflect the intensity and distribution of the signal at the output of the detector and are directly related to final image quality and the dose of the patient [1,2].

The most important imaging parameters are: (α) Quantum Detection Efficiency (QDE), which gives the fraction of the incident radiation which is absorbed in the detector (b) The Modulation Transfer Function (MTF), which describes the change of the incident signal modulation as a function of spatial frequency. It is directly associated with the imaging resolution of the detector, (c) the Noise Power Spectrum (NPS) and the Noise Transfer Function (NTF) which describes the noise (statistical and deterministic) properties of the detection stages and their contribution to the final image (d) Detective Quantum Efficiency (DQE), which describes the ability of the system to transfer the Signal to Noise Ratio (SNR) and (e) Information Capacity (IC), which describes the total diagnostic information that can be detected in bits/mm2. In the case where the detector works in indirect mode that is a luminescent material (scintillator) is attached to a digital photoreceptor (CCD, CMOS, a-Si) the properties of both affect the final image. For the case of the scintillator, except the QDE, important properties are: (i) X-ray to light conversion efficiency, which describes the efficiency the X-rays are transformed into optical photons and is associated with the type, concentration and energy levels of the activator (ii) Optical photon transmission efficiency and distribution to the output, which is affected by the scintillation physical properties and the wavelength of the emitted light and (iii) spectral sensitivity which describes the percentage of the optical photon energy that is actually detected. The last two parameter are directly associated with patient X-ray dose. The photoreceptor inherent properties affecting the imaging output of the detector are: (i) integral nonlinearity, giving the extend where the receptor response is linear with respect to optical photon fluence (ii) Linear and dynamic range, giving the range of the useful detector response (iii) dark current, giving the amount of signal from the detectors when no input is present (iv) read noise, which is related to the total electronic noise of the photoreceptor (v) Quantum efficiency and spectral response, which demonstrate the efficiency of optical photon absorption and secondary carriers generation in the photoreceptor [3-5]. The scope of the MISCIRLU project is an in depth analysis of the aforementioned characteristics.

Materials and Methods

1. Theoretical Approach

The theoretical study of the scintillators as candidate detectors fro X-ray medical imaging is performed via analytical expressions describing the essential parameters used for the evaluation of the scintillators. The first parameter to examine is the absolute efficiency defined as the emitted optical photon power over the incident X-ray exposure. Absolute Efficiency can be calculated by the following formula [6-7]

$$AE = \frac{n_C \gamma(E) t_r \mu(E)(1+\rho)e^{-\mu(E)T}}{2(\mu(E)^2-\sigma^2)} \frac{(\mu(E)-\sigma)(1-\beta)e^{-\sigma T} + 2(\sigma+\mu(E)\beta)e^{\mu(E)T} - (\mu(E)+\sigma)(1+\beta)e^{\sigma T}}{(1+\beta)(\rho+\beta)e^{\sigma T} - (1-\beta)(\rho-\beta)e^{-\sigma T}} \quad (1)$$

where μ(E) is the X-ray energy mass absorption coefficient for X-ray energy E, γ(E) is a conversion factor converting energy fluence (W/m2) into exposure rate (mR/s), tr is the transparency of the phosphor screen substrate and T is the surface density of the scintillator. If the energy spectrum of X-rays, f(E) , is to be taken into account, then AE can be calculated by summing over this spectrum, up to the peak energy (kVp) of the X-ray spectrum. The energy knowledge of AE leads to the calculation of other related parameters like the Detector Optical Gain defined as the ratio of the emitted optical photons over the incident X-ray photons [7]. Factors affecting the values of DOG and absolute efficiency are the absorption efficiency of the X-rays which is affected by the Zeff of the scintillator material, the intrinsic conversion efficiency of the material which is affected by the scintillator energy levels and the optical photon escape probability which is affected by the wavelength of the emitted optical photons and the manufacturing characteristics of the scintillator (i.e. grain size, columns).

The second parameter that is examined is the Modulation Transfer Function of the scintillator, showing the extent to which the scintillator detector can dissolve details. Therefore MTF indicates system resolution. MTF is expressed in the spatial frequency domain and can be calculated by the following formula as [5]:

$$MTF(u) = \frac{\sum_E \sum_t f(\mathrm{E})\overline{M}_x(E,t)\overline{m_o}(E)\overline{G(t)MTF}(u,t)}{\sum_E \sum_t f(\mathrm{E})\overline{M}_x(E,t)\overline{m_o}(E)\overline{G(t)MTF}(0,t)} \tag{2}$$

where Mx(E) descibes the Quantum Detection Efficiency, mo is the gain of the screen (i.e. the number of optical quanta produced per absorbed X-ray and G(t)MTF(u,t) equals to

$$\overline{G(t)MTF(u,t)} = \frac{\tau\rho_i(b+\tau\rho_o)e^{bt} + (b-\tau\rho_o)e^{-bt}}{(b+\tau\rho_o)(b+\tau\rho_i)e^{bT} - (b-\tau\rho_o)(b-\tau\rho_i)e^{-bT}} \tag{3}$$

τ expresses light scattering, σ expresses light absorption and ρo/ρi express front and back screen surface reflection [6]. T in the denominator is the total screen thickness.

Another parameter used for the evaluation of a scintillator is noise. Noise is affected by the statistical nature of X-ray absorption, optical photon production and escape. In additon in the case of digitla imaging the electronic noise should also be considered. It can be evaluated through the Noise Power Spectra (NPS). A scintillators NPS in the spatial frequency domain can be calculated as [5].

$$NPS(u) = \sum_E \sum_t \overline{f}(\mathrm{E})\overline{M}_x(E,t)\left[\overline{m_o}(E)\overline{G(t)MTF}(u,t)\right]^2 \tag{4}$$

Both MTF and NPS are affected by the optical photon scatter and absorption process as well as the thickness of the scintillator. The knowledge of the DOG, MTF and NPS can lead to the calculation of the Detective Quantum Efficiency (DQE). DQE shows the combined effect of resolution and noise in the image per spatial frequency. Since noise is more effective propagating than signal with respect to spatial frequency, DQE values are decreased in terms of frequency. DQE can be calculated as

$$DQE(u) = \frac{[MTF(0)\cdot MTF(u)]^2}{f(E)NPS(u)} \tag{5}$$

Where the nominator in (5) expresses the Signal Power Spectrum

2. Monte Carlo Studies

The evaluation of scintillators can also be performed through Monte Carlo simulations. These take into account the X-ray interactions in the scintillator. This is very important for understanding the role of X-ray scatter and K-characteristic photon production in image quality. The optical photons propagation can also be simulated by Monte Carlo methods by use of the Mie scattering theory. The latter is a powerful tool for simulating the effect of the grain size, optical photon wavelength and screen thickness in MTF [8, 9]

3. Experimental Evaluation

Scintillator materials experimental evaluation comprises optical emission measurements (i.e. absolute efficiency) and image quality metrics measurements (i.e. MTF and NPS). The former is performed by irradiating scintillating screens of different thicknesses with X-rays at various tube voltages (from 50 to 140 kV). The incident exposure rate is measured. Light energy fluence are measured by a photomultiplier (EMI 9798 B) with an extended sensitivity S-20 photocathode and enclosed within a bronze light tight chamber. The output current was amplified and finally measured by a vibrating reed (Cary 400) electrometer operated in current mode. An analogue to digital converter was employed to digitize electrometer's output, which was then stored on a computer. Absolute efficiency was then computed from electrometer's output current and dosimeter data by performing conversions and corrections [10]. MTF can be measured by employing the Coltman formula. However in digital images MTF and NPS are measured by employing the IEC method [5].

4. Current Research

The MISCIRLU project has evaluated thus far the effect of the activator material of the scintillator in MTF, DQE and DOG. This is performed by exploiting equations (1-5) to two scintillator of the same host but different activator, that is $Gd_2O_2S:Tb$ and $Gd_2O_2S:Eu$. The parameters (σ, τ, ρ) needed for calculating equation 3 were obtained by comparing the theoretical results to experimental optical emission results. These values are shown in the following table.

Table 1. Optical parameters of the scintillators

scintillator	$Gd_2O_2S:Tb$	$Gd_2O_2S:Eu$
σ (cm2/g)	27	5.5
β	0.03	0.03
τ (cm2/g)	900	183.33
nC	0.17	0.12
Eλ (eV)	2.46	2
as	0.9	1
ρo, ρι	1, 0.9	1, 0.9

In addition the MISCIRLU project has made initial studies regarding the effect of the grain size in MTF as well as the extend of intrinsic conversion non-proportionality.

Results and Discussion

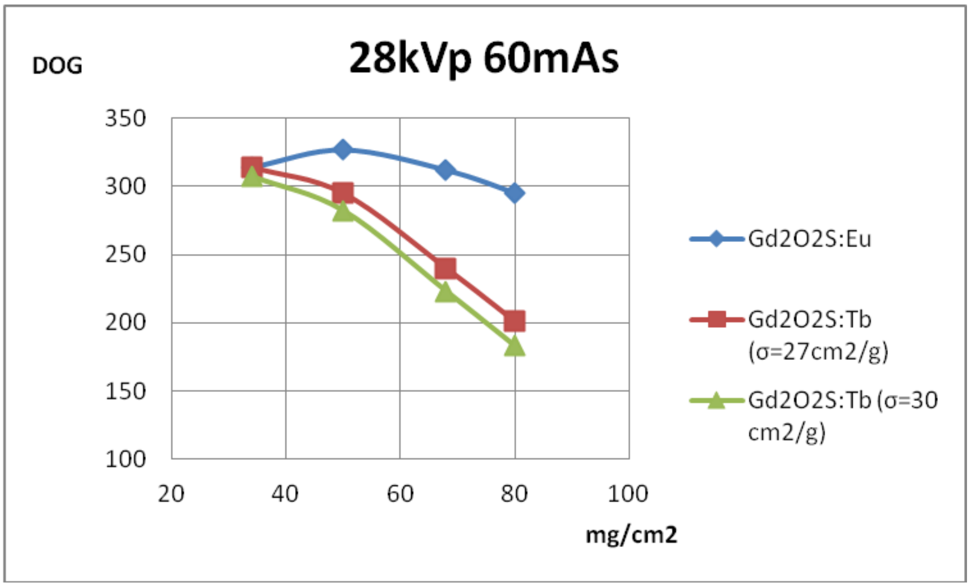

Figure 1. DOG values for Gd_2O_2S:Tb and Gd_2O_2S:Eu scintillators under 28 kVp

In Figure 1 the sensitivity of Gd_2O_2S:Tb and Gd_2O_2S:Eu scintillators is demonstrated described by means of the Detector Optical Gain. It can be observed that Gd_2O_2S:Eu emits a higher number of optical photons per absorbed X-ray due to the lower absorption and scatter (values of σ and τ at Table 1) despite the fact that Gd_2O_2S:Tb produces a higher number of optical photons per absorbed X-ray (values of n_C at Table 1).

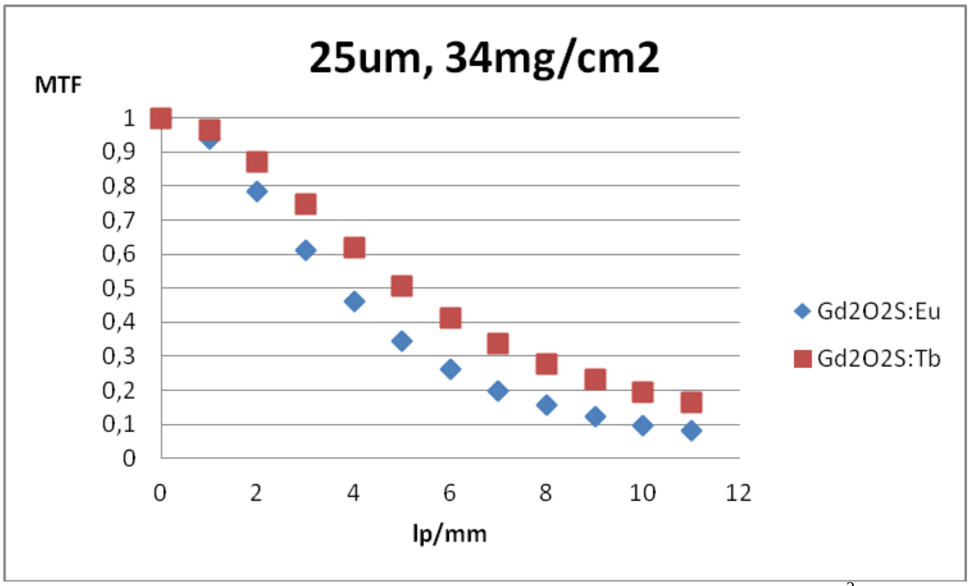

Figure 2. MTF values for Gd_2O_2S:Tb and Gd_2O_2S:Eu scintillators of $34mg/cm^2$ thickness.

In Figure 2 the MTF of Gd_2O_2S:Tb and Gd_2O_2S:Eu scintillators is demonstrated, assuming they are deposited on a digital detector with pixel size 25μm. It can be observed that Gd_2O_2S:Tb has higher MTF values per spatial frequency than Gd_2O_2S:Eu for the specific screen surface density. This may be explained by the fact that the higher optical photon scatter and absorption leads to narrower light bursts in the screen output thus better MTF values.

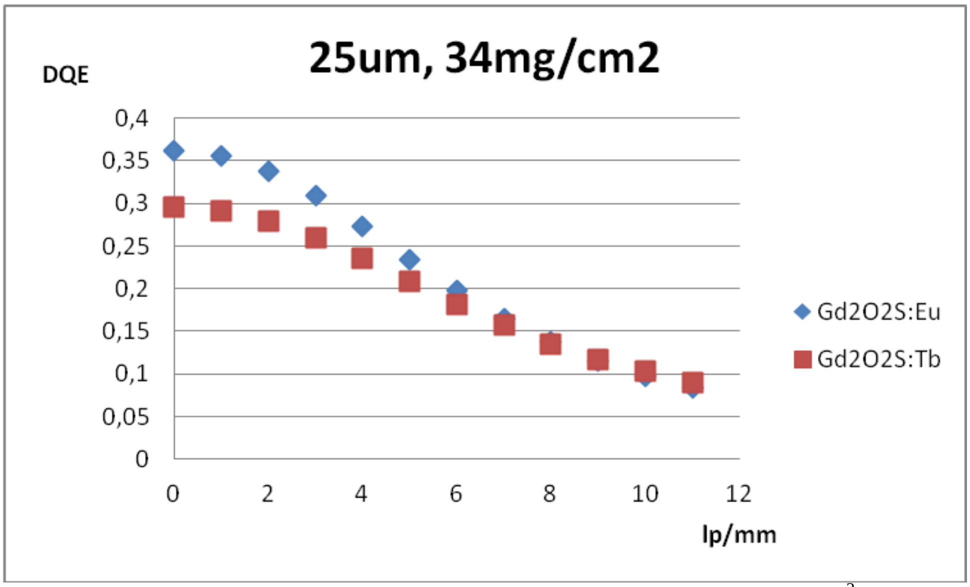

Figure 3. DQE values for Gd_2O_2S:Tb and Gd_2O_2S:Eu scintillators of 34mg/cm^2 thickness.

In Figure 3 the DQE of Gd_2O_2S:Tb and Gd_2O_2S:Eu scintillators is demonstrated, assuming deposited on a digital detector with pixel size 25μm. It can be observed that Gd_2O_2S:Tb has higher MTF values for spatial frequency larger than 8 lp/mm while Gd_2O_2S:Eu is better for smaller frequencies. This result implies that the Signal to Noise Transfer of small details is better transferred with Gd_2O_2S:Tb than Gd_2O_2S:Eu.

With regards to the grain size influence on MTF Monte Carlo simulations have been performed for sizes of 4μm, 6μm, 8μm, 10μm and 12μm, for a scintillator material emitting at 545nm. It was found that 4μm grain size scintillator exhibits higher MTF values. Initial results regarding the non-proportionality effect demonstrated that this effect may be of importance on thick scintillator detectors where an increase in X-ray energy absorption is observed.

Finally the presence of a digital component following the scintillator leads to additional noise source that further reduces the calculated DQE of the scintillator.

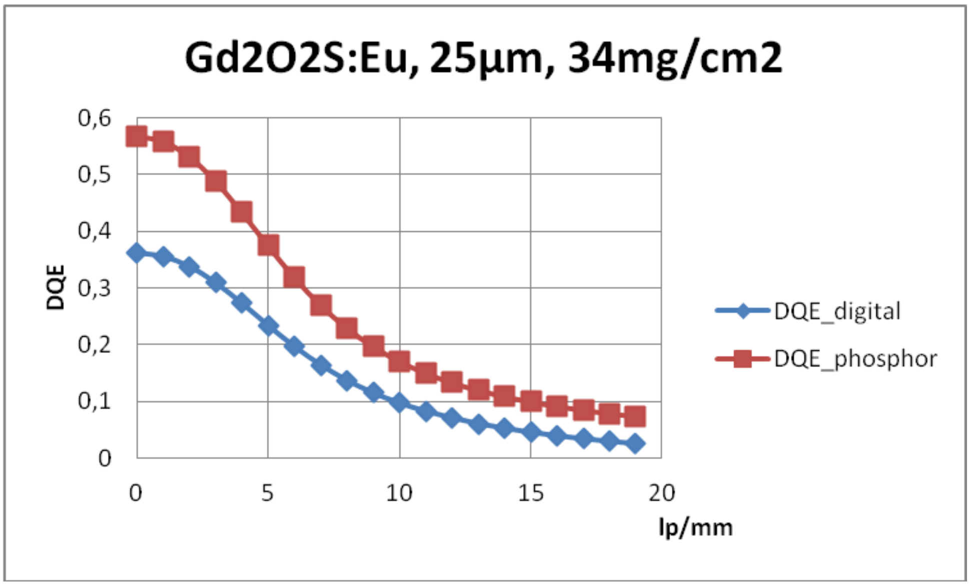

Figure 4. DQE values for $Gd_2O_2S:Eu$ scintillator alone and coupled to a CMOS detector.

This is demonstrated in figure 4 where the presence of the photodetector reduces the DQE to 50% of its zero frequency value. In addition if a direct comparison is performed with experimental data it will be observed that the bit assignment and signal processing performed by the digital detector software further alters the Noise and DQE properties of the scintillator photodetector combination.

Conclusion

The MISCILRU project scope is to examine in deep the ability of a scintillator to be used as an X-ray detector for medical imaging applications. Initial results demonstrate that the choice of the activator, the size of the phosphor grains, the thickness of the detector affects the sensitivity, the detective quantum efficiency the modulation transfer function and intrinsic conversion non-proportionality.

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