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Evaluation of the imaging performance of LSO powder scintillator for use in X-ray mammography

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Abstract

The aim of the present study was to evaluate the imaging performance of Lu_2SiO_5:Ce (LSO:Ce) powder scintillator for use in X-ray detectors used in mammography. LSO:Ce scintillator is a high efficiency, fast emitting material, which in single-crystal form is used in positron emission tomography detectors. A scintillating screen, with a coating thickness of 25 mg/cm^2, was prepared in our laboratory from commercially available LSO:Ce powder (Phosphor Technology Ltd.). The imaging performance of the screen was assessed by experimental determination of the modulation transfer function (MTF) and the noise transfer function (NTF). Experimental MTF values were compared to data obtained by a custom Monte Carlo simulation program. Screen irradiation was performed under exposure conditions employed in mammographic applications (27 kV$_p$, 63 mA$_s$). MTF was determined by the Square Wave Response Function (SWRF) method whereas NTF was estimated by Noise Power Spectrum (NPS) measurements, under uniform screen irradiation. Our results showed that LSO:Ce exhibits high MTF, which is comparable to that of the commercially used Gd_2O_2S:Tb powder scintillator. Considering our MTF results and the fast response of LSO:Ce scintillator screen (40 ns), this material can be considered for use in X-ray mammographic detectors.

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1. Introduction

Cerium (Ce^{3+})-doped scintillators or phosphors are of particular interest in medical imaging because of their very fast response. Cerium-doped Lutetium oxyorthosilicate (LSO) has attracted a great deal of attention due to its many important advantages, such as high luminescence efficiency, high density, short decay time, suitable emission wavelength and very good chemical stability compared to other scintillators [1–3]. The purpose of this study was to evaluate the imaging performance of the powder LSO for use in X-ray mammography. For this reason, LSO powder scintillator was fashioned into a screen with coating thickness of 25 mg/cm^2 prepared in our laboratory. In a previous study, parameters related to the luminescence efficiency, emission spectrum and spectral compatibility of the LSO powder phosphor were investigated [4].

In this communication, frequency domain-related parameters of the LSO, such as the modulation transfer function (MTF) and the noise transfer function (NTF), were investigated. MTF measurements were as follows: (1) reflection mode, where light emitted by the irradiated screen side was measured (front-screen configuration), and (2) transmission mode, where the non-irradiated screen side's light was measured (back-screen configuration). Experimental MTF values were compared with those of Monte Carlo (MC) simulation. The custom-developed MC model was based on the Mie light scattering theory [5,6]. To our knowledge, the imaging performance of powder

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LSO has not been studied under mammographic exposure conditions.

2. Materials and methods

A scintillating screen was prepared in our laboratory by sedimentation of the LSO phosphor powder on a fused silica substrate (spectrosil B). LSO was purchased from Phosphor Technology Ltd. (UK, code: ZBK58/N-S1) with mean grain size of approximately 8 μm. Sodium orthosilicate (Na_2SiO_3) was used as binding material between the powder grains [7,8]. The phosphor was used in the form of a thin layer (test screen) with coating weight of 25 mg/cm^2 for approximating the thickness of screens employed in X-ray mammography. The screen was brought in close contact with a radiographic film (Kodak T-Mat), enclosed in a light tight cassette. The film-screen combination was irradiated by X-rays on a General Electric Senographe DMR Plus mammographic unit (molybdenum anode–molybdenum filter). The X-ray beam was filtered by a 35-mm-thick block of Perspex to simulate beam hardening by human breast. The exposure conditions employed in the experiments were 27 kV$_p$ and 63 mAs.

MTF was experimentally determined by the Square Wave Response Function (SWRF) method using a Nuclear Associates resolution test pattern (typ-53, Nuclear Associates) [9]. Films were developed in an Agfa Scopix LR 5200 film processor. Pattern images, obtained on the films, were digitized by an Agfa Duoscan scanner with scanning parameters 1000 dpi, 8 bit. MTFs were finally calculated via the SWRF in directions vertical with respect to the test pattern lines and by employing Coltman's formula [9,10]. The MTF data were corrected by dividing them with the MTF of the scanner and the MTF of the film, both measured in a previous study [11].

The NTF was obtained through the measurement of the Noise Power Spectrum (NPS). Uniform irradiation of the LSO screen–film combination was performed employing the same exposure conditions as those for MTF measurements [12]. The irradiated films were developed and digitized with the same parameters as in the MTF measurements. Six regions of interest of 128×128 pixels were selected and image density profiles along pixel rows were obtained and averaged. The NPS was calculated by Fourier transforming the auto-correlation function of the pixel value variations, obtained from the digitized film. The film's NPS was also measured and subtracted from NPS data to determine the screen NPS. NTF was then calculated as the square root of the NPS normalized to zero spatial frequency.

The MC model was developed by using as input data only physical (complex refractive index, light wavelength) and structural (grain size, packing density) characteristics of the phosphor. The simulation code was based on: (a) the basic X-ray interactions within the phosphor mass and (b) the light interactions described by the Mie scattering algorithm [5]. After X-ray energy deposition within the phosphor screen an amount of light photons is produced. The number of light quanta is given by the following equation [13]:

$$G(E) = \eta_c \frac{E}{E_\lambda} \tag{1}$$

where E is the X-ray energy absorbed, η_c is the intrinsic X-ray to light conversion efficiency of the phosphor, and E_λ is the energy of the light photons. Light is emitted following an isotropic distribution and its propagation within the screen can be described by the interactions of light quanta with the phosphor grains within the framework of Mie scattering. For each interaction light absorption with respect to scattering is given by the following relation:

$$P_{abs} = \frac{m_{abs}}{m_{abs} + m_{sct}} \tag{2}$$

where m_{abs}, m_{sct} correspond to light absorption and light scattering coefficients, respectively, and are given as follows:

$$m_{abs} = V_d A Q_{abs} \quad \text{and} \quad m_{sct} = V_d A Q_{sct} \tag{3}$$

where V_d is the volume density of the phosphor screen, A is the geometrical cross-section of the grain and Q_{abs}, Q_{sct} are the absorption and scattering efficiency factors.

3. Results and discussions

Fig. 1 shows the experimental MTF curve of the LSO screen measured at 27 kV$_p$ in reflection mode compared with the MC data.

The agreement between MC predictions and experimental MTFs was better at low and higher frequencies (agreement $\pm 5\%$), while in the medium frequency range (2.4–8.5 mm^{-1}) the model overestimated the experimental values by 8–15%. These deviations may due to: (a) the estimated uncertainty in experimental measurements and (b) limitations of the Monte Carlo model (e.g. assumption of Poisson distribution for the production of light quanta,

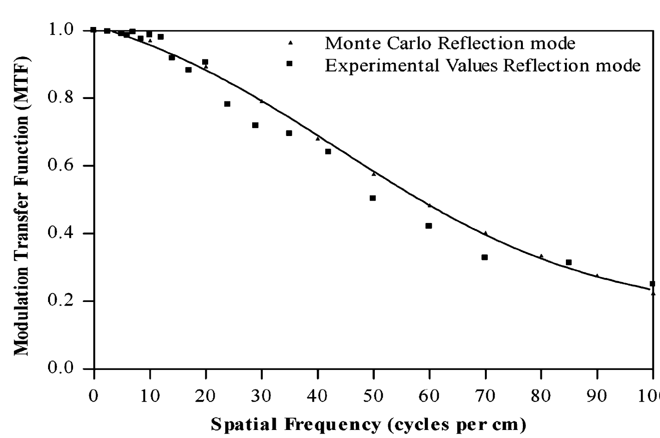

Fig. 1. Comparison of the experimentally determined MTF of the LSO powder screen with the MTF produced by the MC simulation, in reflection mode.

assumption for monochromatic light photons). Curves of similar shape; however, with 20% MTF lower values, were obtained for transmission mode measurements.

Fig. 2 shows a comparison between the MTFs of the $25\,mg/cm^2$ LSO screen and the commercially employed 'Kodak Min-R film-screen system'. The latter is based on a $31.7\,mg/cm^2$ thick Gd_2O_2S:Tb phosphor, exhibiting equal quantum detection efficiency (QDE) with the $25\,mg/cm^2$ LSO screen, used in the present study (see Fig. 3). QDE was calculated considering exponential X-ray absorption. MTF data for Gd_2O_2S:Tb [14] measured at 30 kV, against 27 kV, routinely used in mammography, employed in the present study. As shown in Fig. 2, the LSO MTF curve was found higher than that of the Kodak Min-R.

This difference may be explained by: (i) the lower thickness of the LSO screen; (ii) the lower light emission wavelength of the LSO at 420 nm, i.e. low light wavelength photons are strongly absorbed within the phosphor mass especially in lateral directions, which could lead into a sharp output light distribution, leading to improved resolution (e.g. the scattering efficiency factor for LSO equals to 2.23 compared to 2.03 for the conventional Gd_2O_2S:Tb phosphor material, under identical structural properties); and (iii) the higher value of grain size. However, LSO exhibits lower light emission properties due to (see relation (1)): (a) the intrinsic X-ray to light conversion efficiency of the phosphor (e.g. 0.09 for LSO:Ce [4] compared to 0.15 for Gd_2O_2S:Tb [13] and (b) the energy of light photons which is related to the light wavelength (e.g. 420 nm for LSO:Ce compared to 545 nm for Gd_2O_2S:Tb).

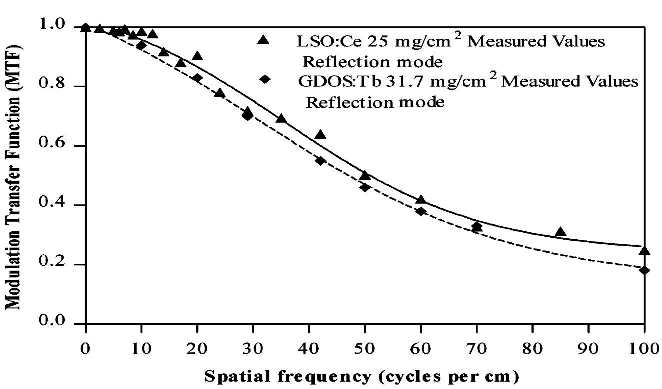

Fig. 2. Comparison of MTFs of LSO and Kodak Min-R screen [14] as measured experimentally in reflection mode.

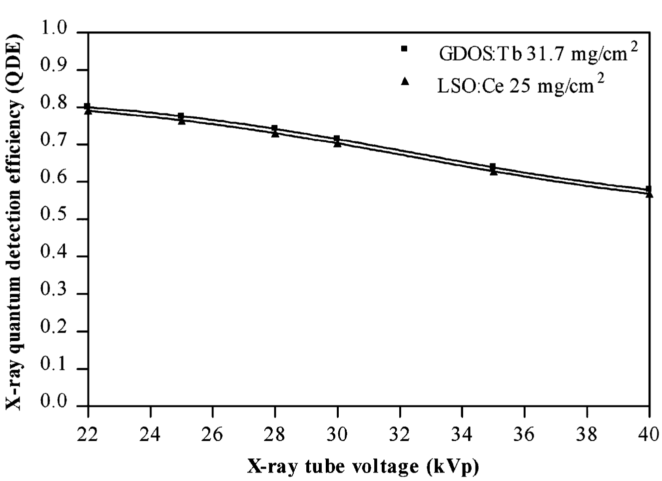

Fig. 3. Variation of calculated QDE of LSO:Ce and Gd_2O_2S:Tb with X-ray tube voltage for 25 and $31.7\,mg/cm^2$ powder screens.

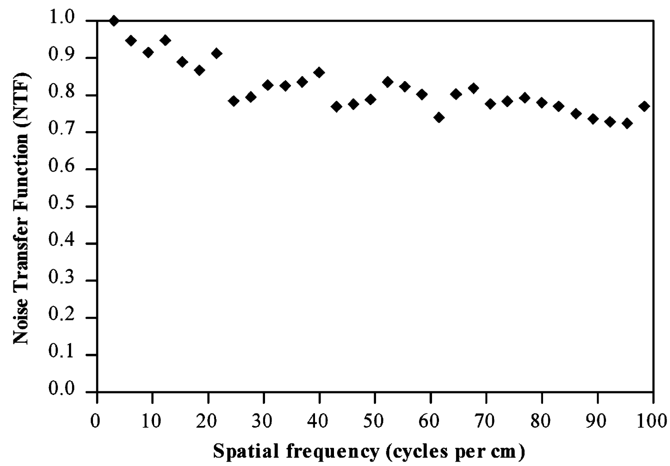

Fig. 4. NTF of the $25\,mg/cm^2$ LSO screen at $27\,kV_p$, 63 mAs, measured in reflection mode.

Fig. 4 shows the measured NTF of the LSO phosphor screen in reflection mode. NTF decreases with spatial frequency, although at a slower rate than MTF. This is because noise is transferred more efficiently than signal in the higher spatial frequencies, as it has been shown in other studies [13,15].

4. Conclusion

In the present study, a LSO:Ce powder scintillator screen of $25\,mg/cm^2$ coating thickness was prepared and examined under X-ray mammographic conditions. Taking into account: (i) its high absorption efficiency at low X-ray energies; (ii) its image quality properties (MTF, NTF); and (iii) its very fast response, this phosphor could be considered for applications in X-ray mammographic imaging systems, both in radiographic cassettes and in digital detectors. This conclusion is reinforced by previous findings showing excellent spectral compatibility with currently used films (0.89–0.96) and adequate compatibility with a-Si-based photodiodes (0.58) [4].

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References

[1] C.L. Melcher, J.S. Schweitzer, Nucl. Instr. and Meth. A 314 (1992) 212.

[2] C.L.Melcher, J.S. Schweitzer, US Patents 4,958,080; 5,025,151; 5,660,627.

[3] C.L. Melcher, M. Schmand, M. Eriksson, IEEE Trans. Nucl. Sci. NS-47 (1992) 965.

[4] S. David, C. Michail, I. Valais, et al., Nucl. Instr. and Meth. A 571 (2006) 346.

[5] P. Liaparinos, I. Kandarakis, D. Cavouras, H. Delis, G. Panayiotakis, Med. Phys. 33 (2006) 4502.

[6] H.C. Van de Hust, Light Scattering by Small Particles, Wiley, New York, 1957.

[7] I. Kandarakis, C. Cavouras, Nucl. Instr. and Meth. A 417 (1998) 86.

[8] I. Kandarakis, D. Cavouras, Radiat. Meas. 39 (2005) 263.

[9] G.T. Barnes, The use of bar pattern test objects in assessing the resolution of screen/film systems, in: A.G. Haus (Ed.), The Physics of Medical Imaging: Recording System Measurements and Techniques, American Association of Physicists in Medicine, New York, 1979, p. 138.

[10] ICRU, Modulation transfer functions of screen–film systems (1986), ICRU Report 41.

[11] I. Kandarakis, D. Cavouras, G.S. Panayiotakis, C.D. Nomicos, Appl. Phys. B 72 (2001) 877.

[12] M.B. Williams, P.U. Simoni, L. Smilowitz, M. Stanton, W. Phillips, A. Stewart, Med. Phys. 26 (1999) 2273.

[13] R.M. Nishikawa, M.J. Yaffe, Med. Phys. 17 (1990) 894.

[14] R.M. Nishikawa, M.J. Yaffe, Proc. SPIE 914 (1988) 128.

[15] R.M. Nishikawa, M.J. Yaffe, Med. Phys. 16 (1989) 773.