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An Investigation of the Imaging Characteristics of the Y_2O_2S:Eu^{3+} Phosphor for Application in X-ray Detectors of Digital Mammography

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Y_2O_2S:Eu laboratory prepared screens were evaluated as mammographic image receptors and were compared to similarly prepared screens of Gd_2O_2S:Tb and Y_2O_2S:Tb phosphor materials, often used in X-ray imaging detectors. The evaluation was performed by determining the Modulation Transfer Function (MTF) and the spatial frequency dependent Detective Quantum Efficiency (DQE). Y_2O_2S:Eu exhibited higher DQE values at low frequencies and given its good spectral matching with digital optical detectors, it may be appropriate for use in X-ray digital mammography. © 1998 Elsevier Science Ltd. All rights reserved

Introduction

Y_2O_2S:Eu is a phosphor material emitting red light due to the presence of Eu^{3+} activator (Tanaka *et al.*, 1976). Initial experimental data on absolute efficiency (emitted light flux over incident X-ray exposure rate) of Y_2O_2S:Eu phosphor screens have been obtained in a previous study (Giakoumakis *et al.*, 1993). It has been found that the light producing efficiency of Y_2O_2S:Eu is comparable to that of Y_2O_2S:Tb, often employed in medical imaging (Arnold, 1979), and additionally that Y_2O_2S:Eu exhibits reduced intrinsic light attenuation properties. However, to our knowledge, Y_2O_2S:Eu has never been used in X-ray image detectors and its image quality characteristics have not been studied.

The low light attenuation properties of Y_2O_2S:Eu indicate that light produced within this phosphor material may be easier transmitted to the phosphor's emissive surface and, thus, be efficiently captured by films, photodiodes or other means of optical detectors. Additionally, the emission spectrum of the Y_2O_2S:Eu red light is better matched, as compared to the green–blue light of Y_2O_2S:Tb, to the spectral sensitivity of photodiodes or CCD arrays employed in digital imaging systems. In the present work the imaging performance of the Y_2O_2S:Eu phosphor was studied under mammography conditions by determining the following parameters: (1) the modulation transfer function (MTF), which expresses the image contrast variation with spatial frequency and the spatial resolution of image detectors; and (2) the detective quantum efficiency (DQE), which describes the signal to noise ratio (SNR) transfer efficiency from the input to the output of an image detector. DQE is related to the patient radiation dose level required to obtain an adequate signal to noise ratio in the final diagnostic image.

Materials and Methods

The phosphor materials (Y_2O_2S:Eu, Y_2O_2S:Tb, Gd_2O_2S:Tb) were supplied in powder form by Derby Luminescents. Test screens of 32 mg/cm^2 and 50 mg/cm^2 were prepared in our laboratory by sedimentation of the phosphor powder on fused silica substrates. Details of the technique followed have been described in previous studies (Giakoumakis *et al.*, 1993; Kandarakis *et al.*, 1996; Cavouras *et al.*, 1996).

The modulation transfer function

The modulation transfer function was experimentally determined employing the square wave response function (SWRF) method (Barnes, 1979). Each screen was used in combination with a film of appropriate spectral sensitivity (Agfa Scopix LT2B for Y_2O_2S:Eu and Agfa Curix Ortho GS for Y_2O_2S:Tb and Gd_2O_2S:Tb) and with a resolution test pattern (typ-53, Nuclear Associates), which comprises line pairs of spatial frequencies ranging from 2.5 to 100 cycles per cm. The pattern–screen–film combination was exposed to 30 kVp molybdenum target X-rays. The resulting pattern images were digitized using a MICROTEC Scanmaker II SP (24-bit color, 1200×1200 dpi) CCD scanner. Low noise SWRFs were obtained by selecting 64 successive image traces transverse to the pattern bars. Trace amplitudes were normalized to the lower pattern frequency. The screen MTF was then calculated from the SWRF values employing Coltman's formula (Barnes, 1979). Results were corrected for screen–film non-linearities, via the H&D characteristic curve, and for the digital acquisiton system MTF (Cavouras *et al.*, 1996). The film's MTF was considered equal to unity for frequencies up to 100 cycles/cm (Beutel *et al.*, 1993).

The screen MTF was also theoretically evaluated (Kandarakis *et al.*, 1997) by modeling the screen to consist of a large number of elementary fluorescent layers of thickness *dt*. The screen was considered to be excited by an incident X-ray fluence, which mathematically may be expressed by a spectral distribution function [S_x in relation (2)]. A fraction η_Q of this fluence is detected by the screen material and the rest is transmitted without interaction. Each absorbed X-ray quantum interacts with the phosphor material at a particular thin layer at depth t from the screen surface and a number (m_0) of optical photons are produced. A fraction G_L of these light photons are transmitted through the screen material and are emitted from the screen output. In fact, G_L is the Fourier transform of the output light photons distribution per absorbed X-ray divided by m_0 and, thus, it depends on spatial frequency ω (Swank, 1973). The number of light photons per absorbed X-ray exciting the screen (m_0G_L) depends on the optical scattering and absorption properties of the phosphor material. The total number of optical photons emitted per unit of screen area was obtained by summing up the light flux contributions from each particular screen layer and integrating over all incident X-ray energies, considering polyenergetic X-ray beams. Thus, the screen MTF was evaluated as:

$$\mathrm{MTF}(\omega) = N_L(\omega)/N_L(0) \tag{1}$$

where N_L (ω) is the spatial frequency dependent mean number of totally emitted optical photons given by:

$$N_L(\omega) = \int_0^{E_0} S_X(E)\eta_Q(E, t_0) \int_0^{t_0} x_R(E, t)\bar{m}_0(E)G_L(\omega, \sigma, \beta, t)\,\mathrm{d}t\mathrm{d}E \tag{2}$$

where, E is the energy of X-ray photons ranging up to E_0 which is the maximum photon energy of the incident X-ray fluence. The latter is given by a molybdenum target X-ray spectrum expressed by the function $S_X(E)$ (Tucker *et al.*, 1991). t_0 is the coating thickness of the screen, $\eta_Q(E,\mathrm{t}_0)$ is the X-ray quantum detection efficiency (QDE) approximated by (Morz and Danos, 1978):

$$\eta_Q(E, t_0) = 1 - \exp[-\mu(E)t_0] \tag{3}$$

where, $\mu(E)$ is the X-ray attenuation coefficient calculated from data published by Storm and Israel (1967). $\bar{\mathrm{m}}_0(\mathrm{E})$ in (2) is the mean number of optical photons created within the phosphor material per X-ray quantum detected by he phosphor screen and it is given by:

$$\bar{m}_0(E) = \frac{\eta_C E}{\int_{E_{\lambda_1}}^{E_{\lambda_2}} S_P(E_\lambda)E_\lambda \mathrm{d}E \Big/ \int_{E_{\lambda_1}}^{E_{\lambda_2}} S_P(E_\lambda)\mathrm{d}E_\lambda} \tag{4}$$

where η_C is the intrinsic X-ray to light conversion efficiency of the phosphor material, $S_P(E_\lambda)$ is the spectrum of the emitted light, which was measured in our laboratory by an Oriel 7240 grating monochromator, E_λ is the energy of the optical photons. λ_1 and λ_2 are the lower and upper wavelengths of the spectrum, respectively. The denominator in (4) is equal to the mean energy of the emitted optical quanta.

The function G_L (ω, σ, β, t) in (2) expresses the fraction of optical quanta, created by one X-ray quantum absorbed by a thin layer at depth t, being transmitted through the phosphor material and emerging at the screen 's emissive surface. G_L (ω, σ, β, t) is a spatial frequency (ω) dependent function that can be calculated (see Appendix) if the values of optical parameters σ and β, related to optical scattering and absorption, are known. $x_R(E, t)$ is a function representing a weighing factor and accounting for the relative X-ray absorption by a thin layer at depth t, given by

$$x_R(E, t) = \mu(E)\exp[-\mu(E)t]\mathrm{d}t \Big/ \int_0^{t_0} \mu(E)\exp[-\mu(E)t]\mathrm{d}t \tag{5}$$

For incident X-ray energies exceeding the K-absorption edge of yttrium at 17 keV, K_α and K_β characteristic radiation is produced due to the K-fluorescence effect. The sites within the phosphor material, where these K-characteristic photons are re-absorbed, become sources of additional optical photon generation. These photons degrade spatial resolution and must be considered in the calculation of the screen MTF. Hence, the relative probabilities of generation and re-absorption of K-characteristic X-rays within the phosphor material were computed in accordance with Chan and Doi (1983). These probabilities were used as weighting factors to calculate the screen MTF as a weighted sum (Que and Rowlands, 1995) of the three MTFs, corresponding to the three possibilities of X-ray absorption: (1) total absorption of incident X-rays, excluding the possibility of energy transferred to K_α or K_β X-rays; (2) absorption of K_α; or (3) of K_β generated X-rays.

By selecting appropriate initial values for parameters σ and η_C and taking the value of β from a previous study (Giakoumakis *et al.*, 1993), relation (1) was used to fit the experimental MTF, employing the Levenberg–Marquard method (Press *et al.*, 1990). Satisfactory agreement was obtained for η_C and σ values (σ = 17.4 cm ^{2}/g, η_C=0.115).

The detective quantum efficiency

The detective quentum efficiency is defined as (Dainty and Shaw, 1974):

$$\mathrm{DQE} = \frac{[\mathrm{SNR}_{\mathrm{out}}]^2}{[\mathrm{SNR}_{\mathrm{in}}]^2} \qquad (6)$$

where $\mathrm{SNR}_{\mathrm{out}}$ and $\mathrm{SNR}_{\mathrm{in}}$ are the output and input signal to noise ratios, respectively.

The spatial frequency dependent detective quantum efficiency DQE (ω) has been formulated as a function of MTF and noise power spectrum (NPS) (Shaw, 1963; Shaw and Van Metter, 1984) as follows:

$$\mathrm{DQE}(\omega) = N_X\left[\frac{\mathrm{d}N_{\mathrm{L}}}{\mathrm{d}N_X}\right]^2\left[\frac{\mathrm{MTF}(\omega)^2}{\mathrm{NPS}(\omega)}\right] \qquad (7)$$

where N_X is the mean number of incident X-ray quanta per unit of screen area and N_{L} is the corresponding mean number of emitted optical quanta. NPS has been expressed (Shaw and Van Metter, 1984) as:

$$\mathrm{NPS}(\omega) = \bar{m}_0(E)G_{\mathrm{L}}(0, \sigma, \beta, t_0)N_{\mathrm{L}} \left[1 + \frac{\varepsilon}{\bar{m}_0(E)}\right]\mathrm{MTF}^2(\omega) + N_{\mathrm{L}} \qquad (8)$$

where, ε denotes the excess of the statistical variance in $m_0(E)$ with respect to the Poisson distribution of $m_0(E)$, G_{L} (0, σ, β, t_0) is the total light transmission efficiency of a screen of thickness t_0 at zerio spatial frequency (ω = 0), given by:

$$G_{\mathrm{L}}(0, \sigma, \beta, t_0) = \int_0^{t_0} x_{\mathrm{R}}(E, t)G_{\mathrm{L}}(0, \sigma, \beta, t)\mathrm{d}t \qquad (9)$$

The emitted optical quanta N_{L} are associated with the incident X-ray quanta N_X as follows:

$$N_{\mathrm{L}} = N_x\eta_Q(E, t_0)\bar{m}_0(E)G_{\mathrm{L}}(0, \sigma, \beta, t_0) \qquad (10)$$

From relations (7)–(10) and assuming Poisson distribution (ε = 0) for m_0 (E) (Shaw and Van Metter, 1984; Nishikawa and Yaffe, 1990), and polyenergetic X-rays:

$$\mathrm{DQE}(\omega) = \eta_{\mathrm{Q}}(E_0, t_0)\left[\frac{\bar{m}(E_0)}{\dfrac{1}{\mathrm{MTF}^2(\omega)} + \bar{m}(E_0)}\right] \qquad (11)$$

where $\bar{m}(E_0)$ is the mean number of light quanta emitted per X-ray quantum absorbed in the screen, averaged over the X-ray spectrum $\bar{m}(E_0)$ was calculated as follows:

$$\bar{m}(E_0) = \int_0^{E_0} S_X(E)\bar{m}(E)\mathrm{d}E \Big/ \int_0^{E_0} S_X(E)\mathrm{d}E \qquad (12)$$

where,

$$\bar{m}(E) = \bar{m}_0(E)\int_0^{t_0} x_{\mathrm{R}}(E, t)G_{\mathrm{L}}(0, t, \sigma, \beta)\mathrm{d}t \qquad (13)$$

is the mean number of light quanta emitted per X-ray absorbed for monoenergetic X-ray beams.

Finally, η_{Q} (E_0, t_0) in (11) is the quantum detection efficiency averaged over the X-ray spectrum as follows:

$$\eta_{\mathrm{Q}}(E_0, t_0) = \int_0^{E_0} S_X(E)[1 - \exp(-\mu(E)t_0]\mathrm{d}E \Big/ \int_0^{E_0} S_X(E)\mathrm{d}E \qquad (14)$$

Results and Discussion

In Fig. 1 and 2 the MTFs of Y_2O_2S:Eu screens are compared to the MTFs of Y_2O_2S:Tb and Gd_2O_2S:Tb screens of approximately equal coating weight. Figure 1 shows the experimental MTF data obtained for the 32 mg/cm^2 screens and Fig. 2 for the 50 mg/cm^2 screens. All screens were prepared and measured under the same conditions. In both figures the MTF of the Gd_2O_2S:Tb screen was better than the corresponding MTFs of the yttrium based phosphors, thus expressing the high detail visibility obtained using Gd_2O_2S:Tb screens. This may be explained because the volume density of Y_2O_2S is about half that of Gd_2O_2S (Zweig and Zweig, 1983). Thus, for the same coating weight Y_2O_2S screens are thicker causing wider spread to light photons traveling towards the screen output, resulting in wider point spread function (PSF), and worse spatial resolution. Additionally, since light attenuation in Gd_2O_2S:Tb is higher (σ = 30 cm^2/g) than in Y_2O_2S:Eu (σ = 17.4 cm^2/g), the wide angled traveling photons in Gd_2O_2S:Tb are greatly attenuated, thus, further limiting the spread of output light photons. The lower light attenuation in Y_2O_2S:Eu is in accordance with the fact that optical scattering is less significant for the longer wavelengths of the red light emitted by this phosphor.

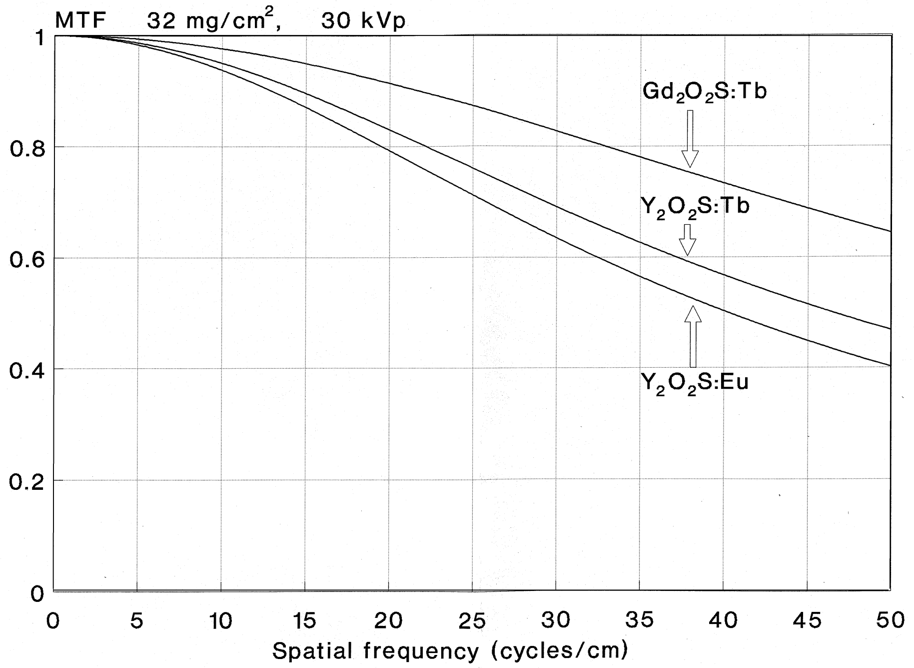

Fig. 1. MTF of Y_2O_2S:Eu, Y_2O_2S:Tb and Gd_2O_2S:Tb 32 mg/cm^2 phosphor screens.

Figures 3 and 4 show the results obtained for the spatial frequency dependent detective quantum efficiency using experimental MTF values. For the range of spatial frequencies up to 22 cycles/cm for the 32 mg/cm^2 screens and up to 17 cycles/cm for the 50 mg/cm^2 screens, Y_2O_2S:Eu exhibited higher DQE and Gd_2O_2S:Tb, while the DQE of Y_2O_2S:Eu was better than that of Y_2O_2S:Tb in the whole frequency range. The rather high DQE values of the europium phosphor at low frequencies must be attributed to the lower light attenuation (σ) within the screen material, which reduces optical photon losses and quantum noise. This contributes to the improvement of SNR_{out} and DQE. In the high frequency range the DQE of Gd_2O_2S:Tb attains higher values than the DQEs of the yttrium phosphors due to the increased differences amongst the corresponding MTFs [see relation (11) and Figs 1 and 2].

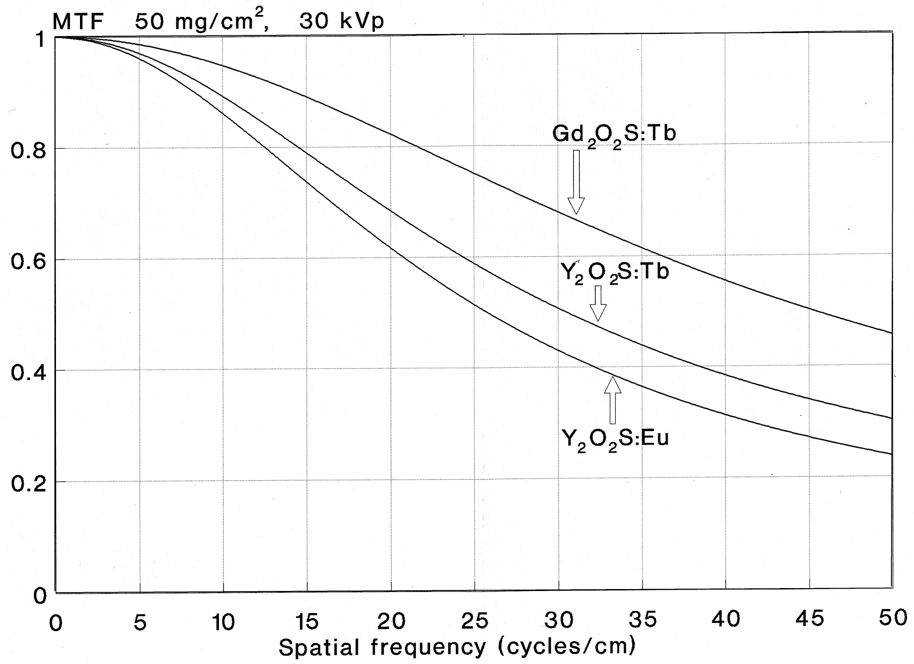

Fig. 2. MTF of Y_2O_2S:Eu, Y_2O_2S:Tb and Gd_2O_2S:Tb 50 mg/cm^2 phosphor screens.

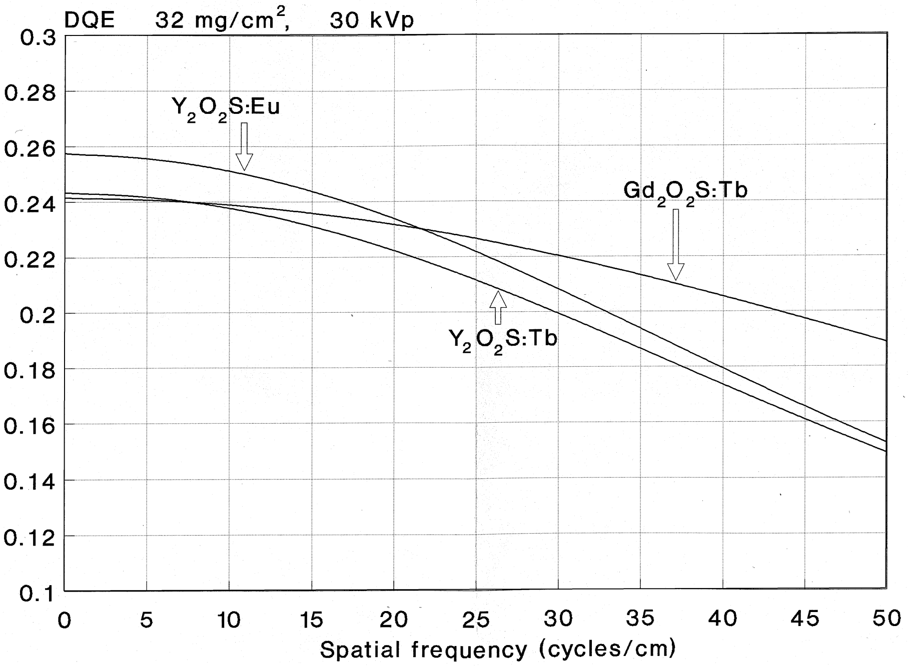

Fig. 3. Spatial frequency dependent DQE of Y_2O_2S:Eu, Y_2O_2S:Tb and Gd_2O_2S:Tb 32 mg/cm^2 screens.

The variation of the calculated zero spatial frequency DQE with screen coating weight is presented in Fig. 5. As expected from Figs 3 and 4, the DQE(0) of the Y_2O_2S:Eu screens is notably higher than the two other materials. The difference in DQE(0) between Y_2O_2S:Eu and the terbium activated phosphors increases with coating weight due to the lower light attenuation properties of the europium activated phosphor.

Conclusively, the Y_2O_2S:Eu screens exhibit better signal to noise ratio than Gd_2O_2S:Tb and Y_2O_2S:Tb in the low frequency range but lower spatial resolution. These results may be of value in the design of detectors employed in digital imaging systems, since in the latter the main purpose is to optimize the signal to noise ratio or DQE (Hillen *et al.*, 1987). Additionally, since the red light produced by europium activated phosphors is very well

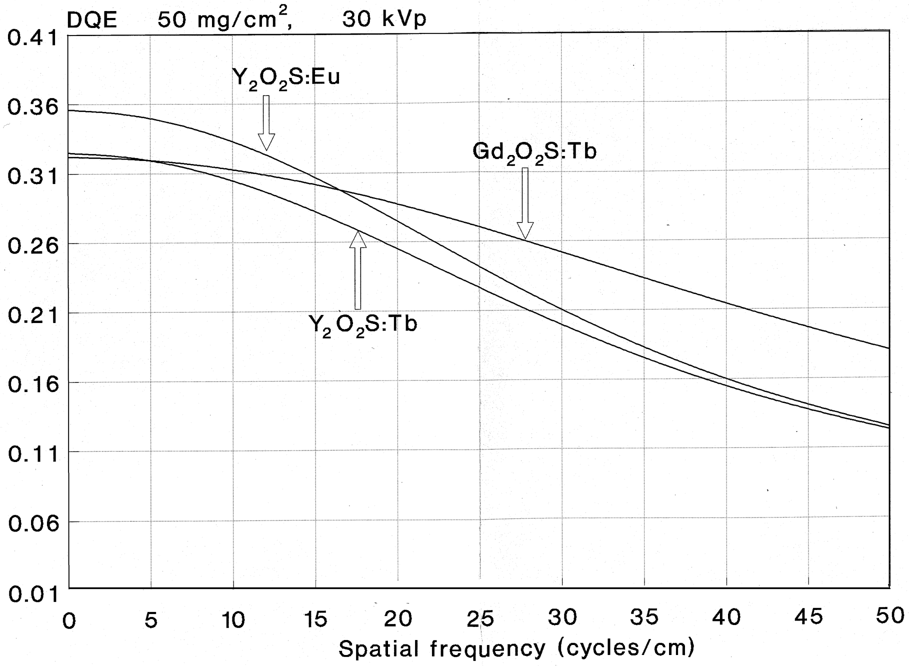

Fig. 4. Spatial frequency dependent DQE of Y_2O_2S:Eu, Y_2O_2S:Tb and Gd_2O_2S:Tb 50 mg/cm^2 screens.

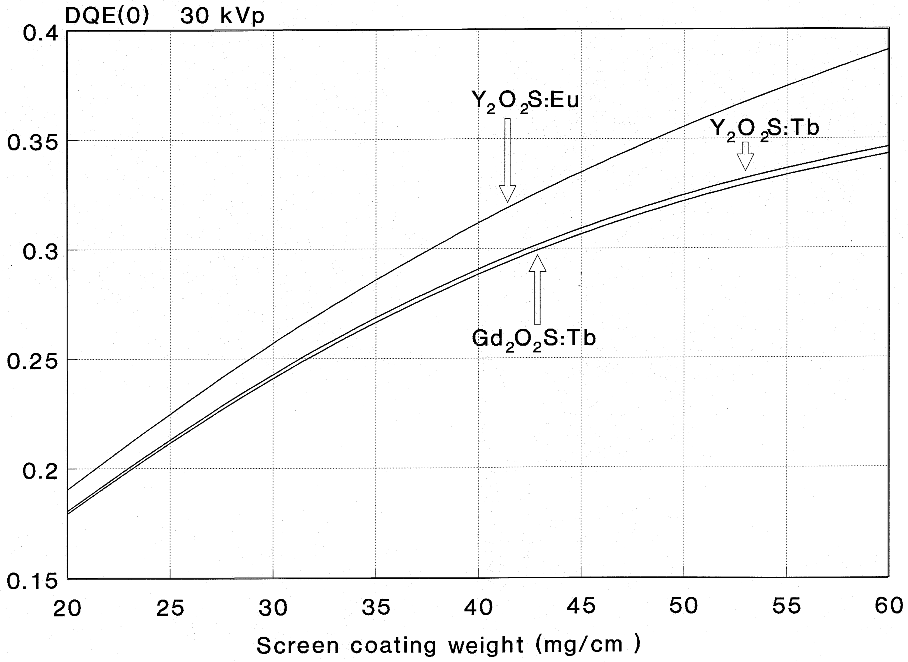

Fig. 5. Variation of zero spatial frequency DQE with screen coating weight.

matched to the sensitivity of digital optical detectors such as Si photodiodes and CCD arrays (Cavouras *et al.*, 1996), Y_2O_2S:Eu could be an appropriate phosphor for use in X-ray digital mammography.

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References

Arnold B. A. (1979) Physical characteristics of screen-film combinations. In *The Physics of Medical Imaging: Recording System, Measurements and Techniques*, ed. Hause A. G., pp. 30–71. American Association of Physicists in Medicine, New York.

Barnes G. T. (1979) The use of bar pattern test objects in assessing the resolution of film/screen systems. In *The Physics of Medical Imaging Recording System Measurements and Techniques*, ed. Hause A. G., pp. 138–151. American Association of Physicists in Medicine, New York.

Beutel J., Apple B. A. and Shaw R. (1993) The role of screen parameters and print-through in the performance of film/screen systems. *Phys. Med. Biol.* **38,** 1181.

Cavouras D., Kandarakis I., Panayiotakis G., Evangelou E. K. and Nomicos C. D. (1996) An evaluation of the Y_2O_2:Eu^{3+} scintillator for application in medical X-ray detectors and image receptors. *Med. Phys.* **23,** 1965.

Chan H. P. and Doi K. (1983) Energy and angular dependence of X-ray absorption and its effect on radiographic response in screen–film systems. *Phys. Med. Biol.* **28,** 565.

Dainty J. C. and Shaw R. (1974) Detective quantum efficiency, signal to noise ratio, and the noise-equivalent number of quanta. In *Image Science*, pp. 152–188. Academic Press, New York.

Giakoumakis G. E., Nomicos C. D., Skountzos P., Koutroubas S., Zisos A., Yiakoumakis E. N., Katsorioti M. C., Kaliakatsos J. A., Rovithi M., Panayiotakis G. S. and Evangelou E. K. (1993) Y_2O_2S:Eu phosphor screens evaluation. *Med. Phys.* **20,** 79.

Hillen W., Schiebel U. and Zaengel T. (1987) Imaging performance of a digital storage phosphor system. *Med. Phys.* **14,** 744.

Kandarakis I., Cavouras D., Panayiotakis G., Agelis T., Nomicos C. and Giakoumakis G. (1996) X-ray induced luminescence and spatial resolution of La_2O_2S:Tb phosphor screens. *Phys. Med. Biol.* **41,** 297.

Kandarakis I., Cavouras D., Panayiotakis G. S. and Nomicos C. (1997) Evaluating X-ray detectors for radiographic applications: a comparison of ZnSCdS:Ag with Gd_2O_2S:Tb and Y_2O_2S:Tb screens. *Phys. Med. Biol.* **42,** 1351.

Ludwig G. W. (1971) X-ray efficiency of powder phosphors. *J. Electrochem. Soc.* **118,** 1152.

Motz J. W. and Danos M. (1978) Image information content and patient exposure. *Med. Phys.* **5,** 8.

Nishikawa R. M. and Yaffe M. J. (1990) Model of the spatial-frequency-dependent detective quantum efficiency of phosphor screens. *Med. Phys.* **17,** 894.

Press W. H., Flannery B. P., Teukolsky S. A. and Vetterling W. T. (1990) *Numerical Recipes in C: The Art of Scientific Computing*. Cambridge University Press, Cambridge, pp. 540–547.

Que W. and Rowlands J. A. (1995) X-ray imaging using amorphous selenium: inherent spatial resolution. *Med. Phys.* **22,** 365.

Shaw R. (1963) The equivalent quantum efficiency of the photographic process. *J. Photogr. Sci.* **11,** 199.

Shaw R. and Van Metter R. (1984) An analysis of the fundamental limitations of screen–film systems for X-ray detection. *Proc. SPIE* **454,** 128.

Storm E. and Israel H. (1967) Photon cross-sections from 0.001 to 100 MeV for elements 1 through 100. *Report LA-3753*. Los Alamos Scientific Laboratory, University of California.

Swank R. K. (1973) Calculation of modulation transfer functions of X-ray fluorescent screens. *Appl. Opt.* **12,** 1865.

Tanaka S., Maruyama Y., Kobayashi H. and Sasakura H. (1976) Electroluminescence in rare earth doped Y_2O_3,La_2O_3 and Y_2O_2S powder layers. *J. Electrochem. Soc.* **123,** 1917.

Tucker D. M., Barnes G. T. and Wu X. (1991) Molybdenum target X-ray spectra: a semiempirical model. *Med. Phys.* **18,** 402.

Zweig G. and Zweig D. A. (1983) Radioluminescent imaging: factors affecting total light output. *Proc SPIE* **419,** 297.

Appendix

The problem of light propagation through the phosphor material has been modeled as a diffusion process described by a diffusion differential equation (Swank, 1973). According to the solution of this equation, the fraction of spatial frequency dependent light quanta generated in the phosphor material at depth t and transmitted to the screen output is given as a function of the optical parameters σ, β (Ludwig, 1971; Swank, 1973; Beutel *et al.*, 1993).

$$G_L(\omega, \sigma, \beta, t) = \frac{\sigma\rho_1[(q\beta + \sigma\rho_0)e^{qt} + (q\beta - \sigma\rho_0)e^{-qt}}{(q\beta-\sigma\rho_0)(q\beta-\sigma\rho_1)e^{qt_0} - (q\beta-\sigma\rho_0)(q\beta-\sigma\rho_1)e^{-qt_0}} \quad \text{(A1)}$$

q in relation (A1) is an optical parameter defined as

$$q = \left(\sigma^2 + 4\pi^2\left[\frac{\omega}{d}\right]^2\right)^{\frac{1}{2}} \quad \text{(A2)}$$

where σ is an optical parameter called reciprocal diffusion length given by:

$$\sigma = \sqrt{[a(a + 2s)]} \quad \text{(A3)}$$

a and s being the light absorption and scattering coefficients of the phosphor. ω is the spatial frequency, d is the density of the phosphor coating, ρ_0, ρ_1 are given by:

$$\rho_n = \frac{(1 - r_n)}{(1 + r_n)} \quad n = 0, 1 \quad \text{(A5)}$$

where, r_0 and r_1 are the reflectivities at the film–screen interface and at the screen–substrate interface, respectively. Values for ρ_1 and ρ_2 were taken from previous studies (Giakoumakis *et al.*, 1993; Cavouras *et al.*, 1996).

Finally, β is an optical parameter given by:

$$\beta = \sqrt{[a/(a + 2s)]} \quad \text{(A6)}$$