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Assessment of the gain transfer function of phosphors for application in medical imaging radiation detectors

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Abstract

Objective: to study various phosphors used in detectors of medical imaging systems by the gain transfer function (GTF), defined in terms of X-ray luminescence efficiency, light spectrum and modulation transfer function.

Materials and methods: four phosphor materials, La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu and Y_2O_3:Eu were used in the form of fluorescent layers prepared in the laboratory. The GTF was determined at 30 kVp and 80 kVp X-ray tube voltages for various phosphor coating weights.

Results: La_2O_2S:Tb, which was the highest density and effective atomic number phosphor used, was found to exhibit the best GTF performance at 80 kVp. At 30 kVp, the yttrium based phosphors were found of increased performance. This is mainly due to the proximity of the X-ray energy to the K-absorption edge of yttrium at 17 keV. Europium activated phosphors were found to perform very well when combined with the red sensitive film and the silicon photodiode.

Conclusion: The GTF may be a useful method for comparing and selecting phosphor materials for use in various medical imaging applications.

Keywords: Phosphors; X-ray luminescence; MTF; Spectral compatibility

1. Introduction

X-ray radiography, fluoroscopy, digital imaging, computed tomography, gamma cameras employ phosphor materials in their detector systems. These materials convert incident X-rays into light, which is then captured by an optical photon detector. The latter produces either an electronic signal (digital radiography, computed tomography, fluoroscopy) or optical density specks in radiographic films. The choice of phosphor material for a specific imaging application is of crucial importance, since both patient dose and image quality may be significantly influenced by the intrinsic physical properties of the material, namely the effective atomic number, the density, and the type of activator. Previous studies in evaluating the useful signal produced by phosphor materials have mainly relied on the determination of the efficiency (sensitivity) and/or the modulation transfer function (MTF) [1–9] of phosphor screens, screen-film combinations, image intensifiers or other type of imaging detectors.

In this study a method is presented for evaluating the performance of phosphor materials by a single quantity, the gain transfer function (GTF). GTF is defined as a combination of the phosphor efficiency (gain), the MTF, and the spectral compatibility of the phosphor's light with optical detectors. The method was used to compare the performance of various phosphor materials that differ in their intrinsic properties, i.e. effective atomic number, density and type of activator.

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2. Materials and methods

The performance of medical image receptors is often assessed by their transfer characteristic curve. The latter describes the relationship between the input signal (incident radiation) fed into the receptor and the output signal (light, electrons, optical density). The slope at any point of the characteristic curve gives the efficiency of the receptor to produce an output signal for a given input signal. For radiographic screen-film systems, this curve is usually called the Hurter–Driffield curve [1–3]. In the case of simple fluorescent layers, employed in radiographic cassettes or other type of X-ray imaging detectors, the corresponding transfer characteristic curve could be defined as the curve describing the emitted light intensity (I_L) as a function of the incident X-ray energy flux (Φ_Q). The slope of this curve may be defined as the gain (η_G) of the fluorescent layer and is given by the ratio

$$\eta_G = \frac{dI_L(w)}{d\Phi_Q} \tag{1}$$

where w is the coating weight of the fluorescent layer. This ratio has been also defined as the X-ray luminescence efficiency (XLE) of a phosphor screen [4–6]. The emitted light intensity (I_L) depends on the spectral characteristics of the incident X-ray beam and the intrinsic properties of the phosphor material, e.g. effective atomic number, density, activator. I_L also depends on the coating weight w of the layer.

For photographic films and radiographic screen-film combinations, an expression of the slope of the characteristic curve into the spatial frequency domain is often employed. This expression uses the concept of contrast transfer function (CTF), defined as the product of the slope times the MTF [3,7]. Accordingly, for a fluorescent layer of coating weight w, we may define the gain transfer function (G_P), as follows:

$$G_P(u, w) = \left[\frac{dI_L(w)}{d\Phi_Q}\right] M_P(u, w) \tag{2}$$

where u is the spatial frequency and M_P denotes the MTF of the phosphor. In medical imaging, where fluorescent layers are used in combination with optical detectors (films, photocathodes, photodiodes), the spectral matching between the emitted phosphor light and the optical detector sensitivity must be taken into account. This is because the degree of spectral matching affects the amount of light utilized to form the final image. Thus, dI_L in Eqs. (1) and (2) is reduced by a factor c_{SP}, expressing the fraction of emitted light that can be detected by the optical detector, which exhibits a specific spectral distribution of sensitivity. c_{SP} can be calculated by the relation [8,9]

$$c_{SP} = \frac{\int_\lambda \bar{I}_L(\lambda) S_{OD}(\lambda)\, d\lambda}{\int_\lambda S_{OD}(\lambda)\, d\lambda} \tag{3}$$

$\bar{I}_L(\lambda)$ is the normalized spectral distribution of the emitted light intensity, $S_{OD}(\lambda)$ is the normalized spectral sensitivity of the optical detector, and λ is the light wavelength.

By taking into account c_{SP}, we may define the effective gain transfer function as follows:

$$G_P(\Phi_Q, u, w) = \left[\frac{dI_L(w)}{d\Phi_Q}\right] c_{SP} M_P(u, w) \tag{4}$$

The GTF and the effective GTF were determined for a number of phosphor materials, which were used in the form of test phosphor screens prepared in our laboratory with various coating weights. The materials employed were the following: (1) La_2O_2S:Tb, which has been used in radiographic screens and in image intensifiers, (2) Y_2O_2S:Tb, employed in radiographic screens and in digital radiography detectors, (3) Y_2O_3:Eu that has been proposed for use in computed tomography detectors, and (4) Y_2O_2S:Eu which, to our knowledge, has not been used in medical imaging. The selection of these phosphors was based on the necessity to investigate the influence of the intrinsic physical properties (atomic number, density, activator) on phosphor performance; Y_2O_2S:Tb and La_2O_2S:Tb differ mainly in the atomic number, Y_2O_3:Eu density is significantly lower than the rest, and Y_2O_2S:Eu and Y_2O_2S:Tb differ in the type of activator. Tb and Eu cause the emission of green and red light respectively [1,2,10].

The phosphor materials were supplied in powder form with an average grain size of approximately 7 μm. The phosphor layers (screens) were prepared by sedimentation of the powder on fused silica substrates [6,8]. The reflecting properties of the substrate are often expressed by the parameter ρ ($\rho = (1-r)/(1+r)$), where r is the reflectivity of the phosphor–substrate interface) [11]. ρ was found to be equal to 0.90 [8], which is close to that found by others [11] for commercial screens. A mixture consisting of the appropriate amount of phosphor powder, 1 l of deionized water and 2 l of Na_2SiO_3 were employed for the sedimentation process. Na_2SiO_3 was used as binding material between the phosphor grains. The refractive index of this binding material is 1.353, which is slightly lower than the binding refractive indices (1.48–1.51) in common radiographic imaging detectors [12]. The packing density of the phosphor layers was slightly higher than 50% the density of the pure phosphor. This packing density is very close to that of commercially available phosphor based X-ray image receptors. The coating weight of the phosphor layers ranged from approximately 20 to 150 mg cm^{-2}, most often employed in various x-ray imaging techniques.

The phosphor layers were excited to luminescence by X-rays employing tube voltages of 30 kVp (ripple 2%) and 80 kVp (ripple 5%), often used in mammographic applications and in general radiography respectively. The emitted light intensity (I_L) was measured by a photomultiplier (EMI 9558 QB) coupled to an electrometer (Cary 401) as described in previous studies [5,6,8,9]. The incident X-ray flux (Φ_Q) was determined from the incident exposure rate [13,14], which was measured by a PTW dosemeter (type No. 23333). To determine the spectral matching factor (c_{SP} in Eq. (3)), the optical spectrum of the emitted light ($I_L(\lambda)$ in Eq. (3)) was determined by a grating type monochromator (Oriel 7240) and the spectral sensitivity ($S_{OD}(\lambda)$ in Eq. (3)) data for various optical photon detectors were obtained from manufacturer's data. The detectors considered were the following: (1) the Agfa Curix Ortho GS orthochromatic film, sensitive to green light emitted by the terbium (Tb^{3+}) activated phosphor materials, (2) the Agfa Scopix LT 2B laser imager film, which is sensitive to red light and hence it is appropriate for use with the europium (Eu^{3+}) activated phosphors Y_2O_2S:Eu and Y_2O_3:Eu, (3) the silicon (Si) photodiode, which is used in detectors of various digital radiography and computed tomography systems, (4) the extended sensitivity (ES/20), photocathode used in various types of image intensifiers and photomultipliers, employed in X-ray fluoroscopy, in digital imaging, and in γ-ray detectors, respectively.

The MTF was determined by performing measurements according to the square wave response function (SWRF) method [15,16]. The SWRF was first measured using a suitable test pattern (typ-53 of Nuclear Associates) comprising lead line pairs with spatial frequencies ranging from 0.25 to 10 cm^{-1}. This pattern was imaged on an X-ray film, which was illuminated by the light of the X-ray excited phosphor layer. The film was placed in contact with the phosphor's emitting surface while the test pattern was placed on the irradiated phosphor side. The pattern images (SWRF) were digitized on a Microtec Scanmaker II SP (1200 × 1200 dpi) scanner. The MTF was then determined using the digitized SWRF values and Coltman's formula [6,8,15]:

$$M_P(u, w) = \frac{\pi}{4} \sum_{k=1}^{\infty} b_K \frac{\mathrm{SWRF}[(2k-1)u, w]}{(2k-1)},$$

$$\text{for } k = 1, 3, 5, \ldots \tag{5}$$

where $b_K = 0$, for $m < n$

$$b_K = (-1)^n(-1)^{k-1}, \quad \text{for } m = n \tag{6}$$

where n is the number of prime factors other than unity in $(2k-1)$, m is the number of prime factors other than unity which appear only once in $(2k-1)$ [16].

To obtain the phosphor MTF, the values obtained by Eq. (5) were divided by the scanner and film MTFs, which were determined by the same method.

3. Results and discussion

Fig. 1 shows the variation of GTF with spatial frequency for seven La_2O_2S:Tb phosphor layers, measured at 80 kVp. As it can be observed: (1) GTF decreases with increasing frequency and (2) the zero

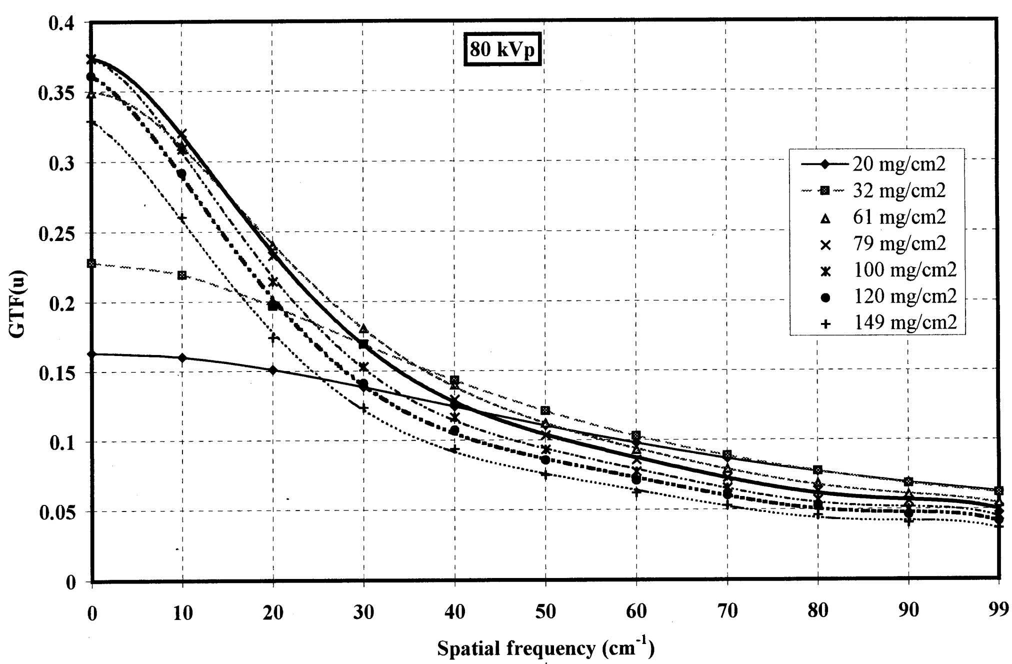

Fig. 1. Variation of GTF with spatial frequency for seven La_2O_2S:Tb phosphor layers, measured at 80 kVp X-ray tube voltage.

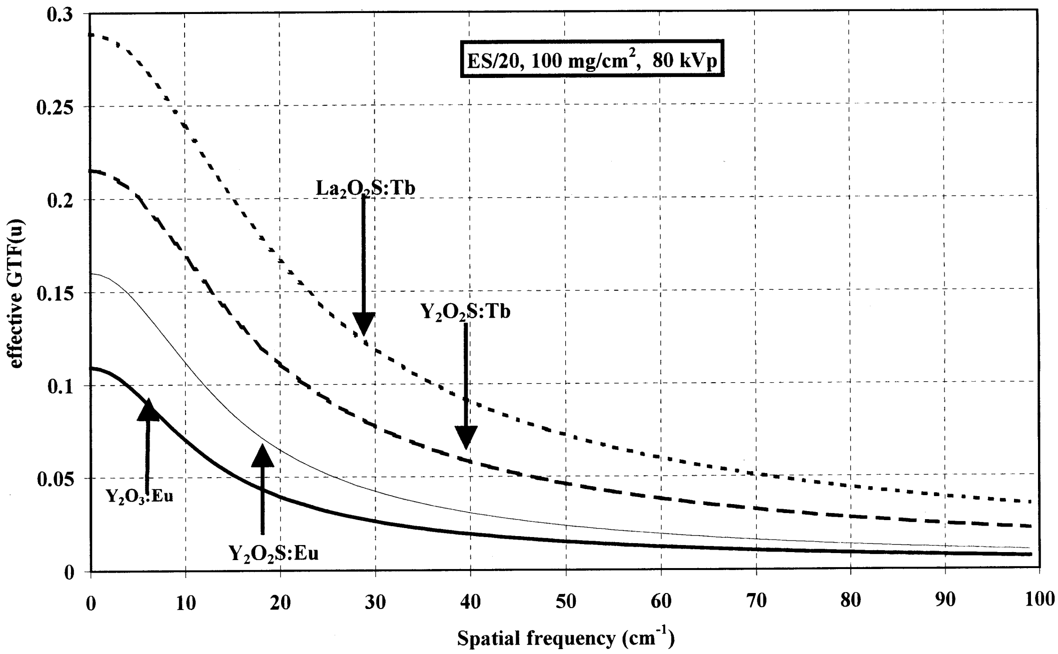

Fig. 2. Variation of effective GTF with spatial frequency measured at 80 kVp for 100 mg cm^{-2} La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor layers combined with the ES/20 photocathode.

frequency GTF value (GTF(0)) and (3) the rate of decrease are different for each coating weight. Regarding GTF(0), which is identical to the phosphor's X-ray luminescence efficiency, it was found maximum for the 79 and 102 mg cm^{-2} layers. GTF(0) depends on the phosphor's X-ray absorption efficiency, which in turn depends on the energy of the X-rays and on the density, effective atomic number, and thickness of the phosphor layer. At the low coating weights of 20 and 32 mg cm^{-2}, the layers are very thin to adequately absorb the incident X-ray quanta. Accordingly, the emitted light intensity is relatively low, giving low GTF(0) values. As coating weight increases, more X-ray quanta are absorbed producing higher quantities of light. This may be evident by comparing the GTF(0) of 79 and 102 mg cm^{-2} with the thinner layers. However, GTF(0) was found lower for the 120 and 149 mg cm^{-2} layers than for the 79 and 102 mg cm^{-2}. This may be explained by considering that in thick phosphor layers, a considerable amount of light is absorbed within the phosphor layer. Regarding the rate of GTF decrease with frequency, it is observed that the GTF(u) of the 20 and 32 mg cm^{-2} thin layers decreases rather slowly with spatial frequency. At frequencies higher than 40 cm^{-1}, the 32 mg cm^{-2} layer attained the highest GTF(u) values while the 20 mg cm^{-2} layer was found equally high after 60 cm^{-1}. These results may be explained by considering the significance of the MTF influence on the GTF at non-zero frequencies. The MTF is mainly determined by the extent of light spread at the output surface of the phosphor. When light spread increases, spatial resolution and hence MTF decrease. If the phosphor is thick, laterally generated and scattered photons are projected and spread over a wide area at the phosphor's output surface. This effect decreases the MTF and thus GTF(u). As frequency increases, the influence of MTF on GTF is more significant than the corresponding influence of X-ray luminescence efficiency. This explains the very low GTF(u) values of the thick layers 149, 120 and 102 mg cm^{-2} in the medium to high frequency region (30–100 cm^{-1}).

Fig. 2 shows the effective GTF(u) curves corresponding to approximately 100 mg cm^{-2} layers prepared from La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor materials, combined with the ES/20 photocathode. Measurements were performed at 80 kVp. These data are useful for fluoroscopy and for digital radiography systems employing image intensifiers. La_2O_2S:Tb and Y_2O_3:Eu phosphors displayed the highest and lowest GTF(u) values respectively. The phosphor material ranking depicted in Fig. 2, may be explained by considering the intrinsic physical properties of the four materials and the c_{SP} factor (see Eqs. (3) and (4)). La_2O_2S:Tb is the phosphor with the highest effective atomic number (atomic number of La: 57) and density (5.5 g cm^{-3}). These two properties augment X-ray absorption efficiency resulting in higher light intensity production and, hence, higher GTF(0) values. Additionally, La_2O_2S:Tb is activated with terbium, which increases the intrinsic X-ray to light conversion efficiency as compared to the intrinsic efficiency of the europium activated phosphors [8]. Also, for a given coating weight, a high density material such as La_2O_2S:Tb will produce thinner layers than a material of lower density. This limits light spread thus increasing MTF, which in turn ameliorates GTF at frequencies higher than zero.

The differences between Y_2O_2S:Tb and Y_2O_2S:Eu are due to the different ion activators (Tb^{3+} or Eu^{3+}) and c_{SP} factors, since their density and effective atomic number are equal (5 g cm^{-3}). The terbium activated phosphor exhibits higher X-ray to light conversion efficiency producing more light. Additionally, the spectrum of this light lies in the green region of the visible spectrum, which is more efficiently detected by the ES/20 photocathode than the red light emitted by Y_2O_2S:Eu. This results in higher c_{SP} factor ($c_{SP} = 0.88$) for Y_2O_2S:Tb. On the other hand, laterally directed green light photons, which travel long trajectories within the phosphor mass to escape the layer, are significantly attenuated due to their higher frequency as compared with red light photons traveling along the same pathways. Thus, light spread at the phosphor's output is restricted in the case of green light and hence MTF is ameliorated. This may explain the superiority of the GTF(u) of Y_2O_2S:Tb. The lowest GTF values of Y_2O_3:Eu are mainly due to its low density (3 g cm^{-3}) and to its slightly lower effective atomic number. These two properties cause a decrease in X-ray absorption. Additionally, for a certain coating weight, a low density phosphor would give thicker layers, and thus lower MTF values.

Fig. 3 shows the GTF variation with spatial frequency of La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu thin layers of approximately 32 mg cm^{-2} combined with the ES/20 photocathode and measured at 30 kVp. It is important to note the GTF superiority of Y_2O_2S:Tb over that of La_2O_2S:Tb. The enhanced performance of the yttrium based phosphor is due to the energy of the K-absorption edge of yttrium (17 keV), which is closer to the mean energy of the 30 kVp beam than that of the lanthanum K-edge (39 keV). However, as spatial frequency increases, the MTF of the La_2O_2S:Tb layer should decrease slower than the MTF of Y_2O_2S:Tb, due to the lower density and hence higher thickness of yttrium based phosphors. This may explain the convergence of GTF values of La_2O_2S:Tb and Y_2O_2S:Tb after 60 cm^{-1}.

Figs. 4 and 5 show effective GTF curves corresponding to phosphor–radiographic film combinations. It is worth noting that at 80 kVp Y_2O_2S:Eu was better than Y_2O_2S:Tb for very low frequencies (Fig. 4). Also, at 30 kVp, the Y_2O_2S:Eu-film and Y_2O_3:Eu-film combinations showed higher effective GTF than La_2O_2S:Tb-film for frequencies up to 45 and 17 cm^{-1}, respectively. This is due to the very high spectral compatibility ($c_{SP} = 0.92$) of the red light emitted by the europium phosphors with the spectral sensitivity of the red sensitive films. In contrast, the corresponding compatibility between the terbium activated phosphors and the green sensitive films is lower ($c_{SP} = 0.7$–0.8).

Figs. 6 and 7 show effective GTF curves corresponding to combinations of the four phosphor materials with the Si photodiode. Fig. 6 depicts data concerning measurements performed at 80 kVp on 100 mg cm^{-2} phosphor layers while Fig. 7 shows data at 30 kVp on

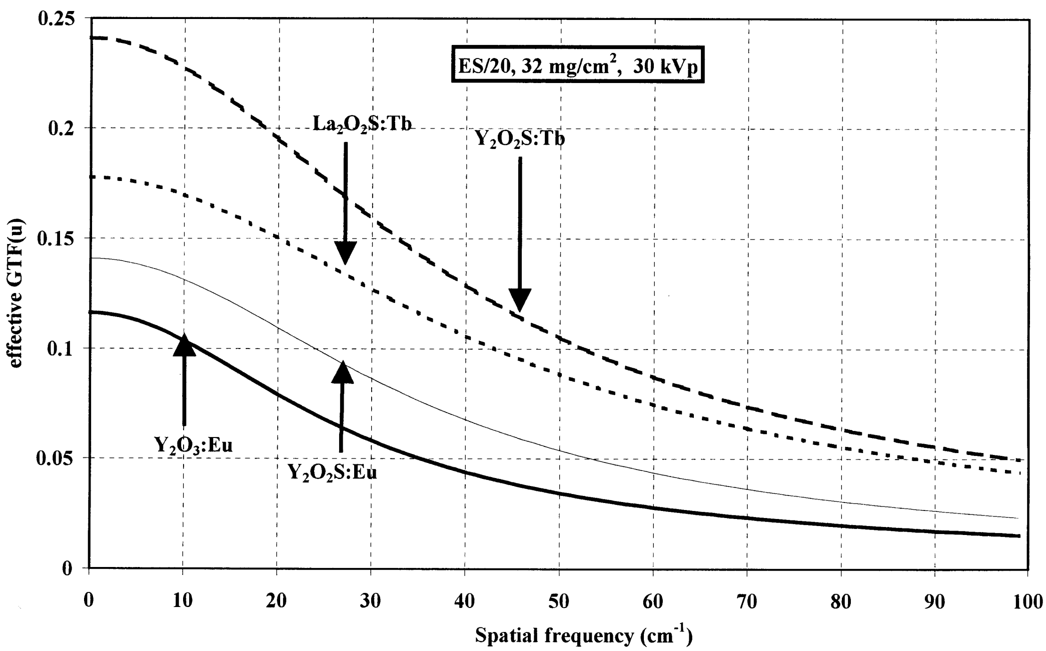

Fig. 3. Variation of effective GTF with spatial frequency measured at 30 kVp for 32 mg cm^{-2} La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor layers combined with the ES/20 photocathode.

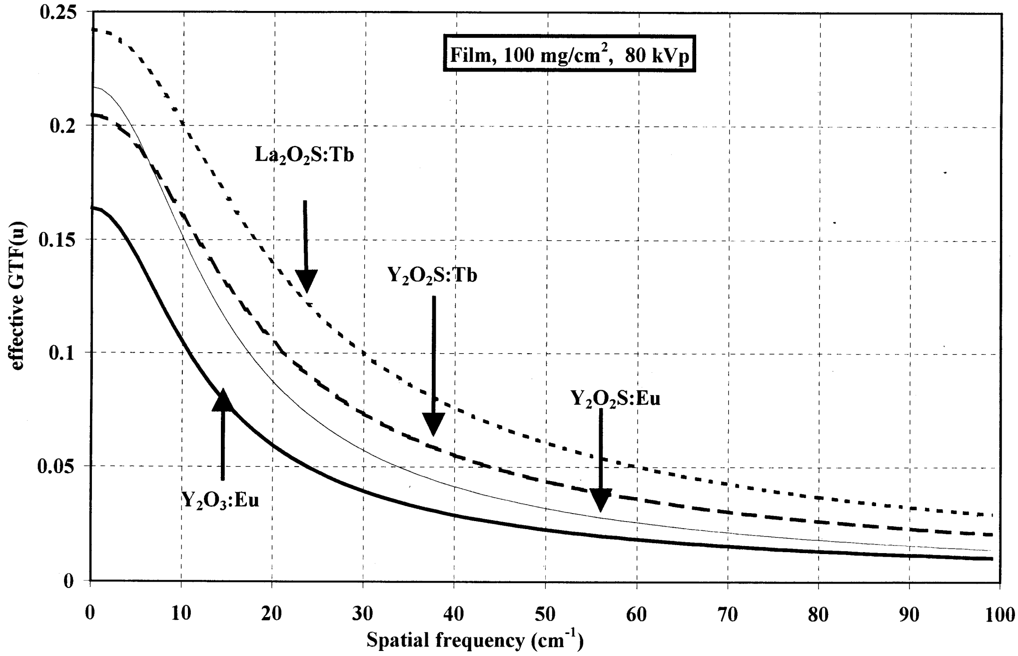

Fig. 4. Variation of effective GTF with spatial frequency measured at 80 kVp for 100 mg cm^{-2} La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor layers combined with the radiographic film.

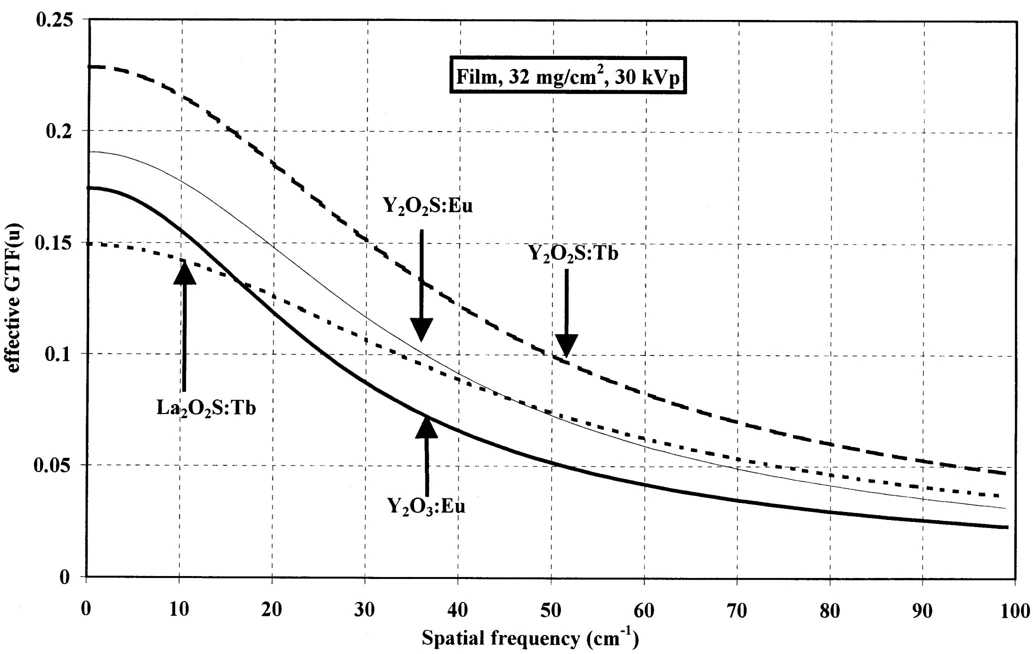

Fig. 5. Variation of effective GTF with spatial frequency measured at 30 kVp for 32 mg cm^{-2} La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor layers combined with the radiographic film.

32 mg cm^{-2} layers. At 80 kVp the La_2O_2S:Tb–Si combination showed highest values. However, it is interesting to note that Y_2O_2S:Eu is clearly better than Y_2O_2S:Tb for frequencies up to 60 cm^{-1}, while Y_2O_3:Eu is better than Y_2O_2S:Tb for very low frequencies. At 30 kVp, the Y_2O_2S:Eu–Si combination is even higher than La_2O_2S:Tb–Si for frequencies up to 25 cm^{-1}. At the very low frequency region, Y_2O_3:Eu was found close to La_2O_2S:Tb and better than Y_2O_2S:Tb. Data concerning europium activated phosphors are affected by the spectral compatibility of the red light with the Si photodiode, which is higher than the corresponding compatibility of the terbium activated phosphors. However, at high frequencies and at high X-ray ener-

gies (80 kVp) La_2O_2S:Tb–Si remains the best combination mainly due to its better MTF. These results are interesting for digital imaging systems and especially the results obtained at 30 kVp are applicable to digital mammography.

In conclusion the effective gain transfer function, expressed in terms of the phosphor's efficiency, the MTF, and the spectral compatibility, was found to vary depending on the intrinsic physical properties of the phosphor materials (density, effective atomic number, activator). High density and high effective atomic number materials show better performance at relatively high X-ray tube voltages. Terbium activated phosphors performed better with currently used photocathodes and

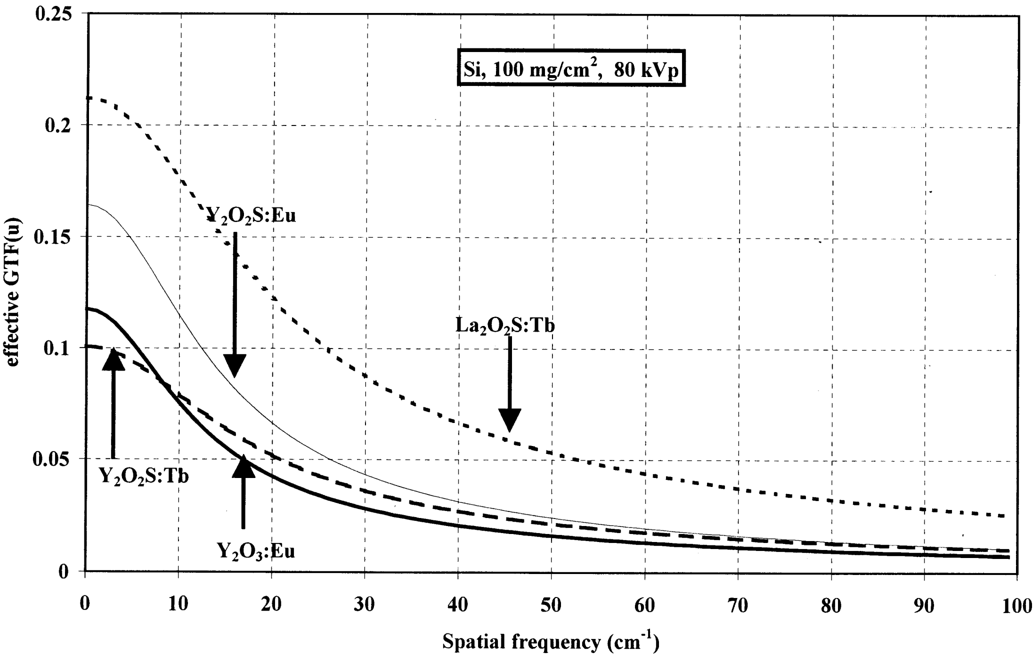

Fig. 6. Variation of effective GTF with spatial frequency measured at 80 kVp for 100 mg cm^{-2} La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor layers combined with the Si photodiode.

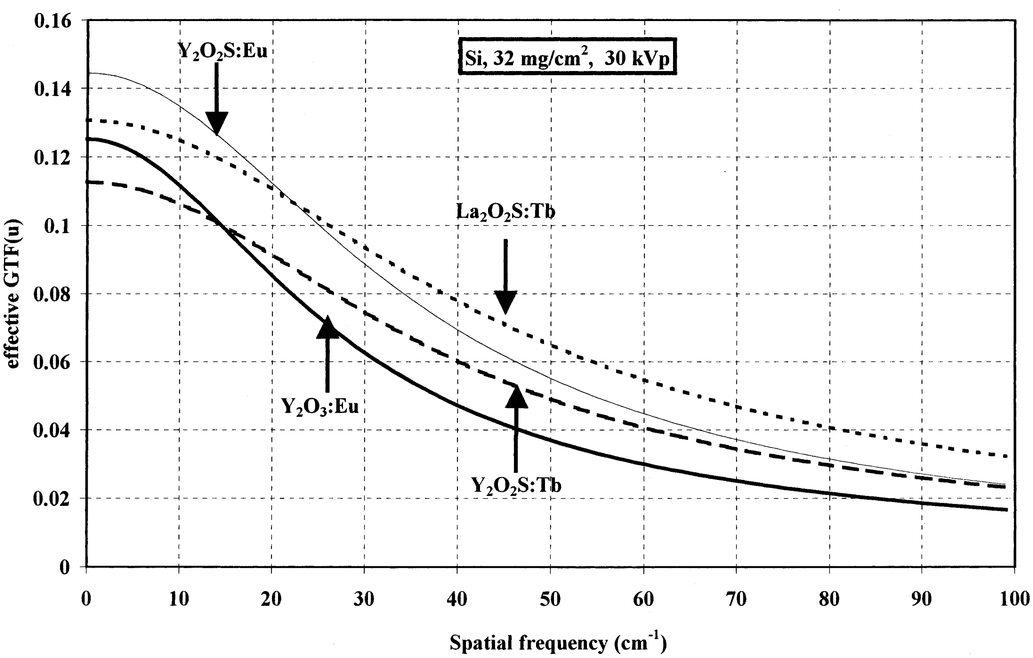

Fig. 7. Variation of effective GTF with spatial frequency measured at 30 kVp for 32 mg cm^{-2} La_2O_2S:Tb, Y_2O_2S:Tb, Y_2O_2S:Eu, and Y_2O_3:Eu phosphor layers combined with the Si photodiode.

green sensitive films. On the other hand, europium activated phosphors were better combined with the Silicon photodiode and the red sensitive films. The gain transfer function may be a useful mean for comparing and selecting phosphor materials for various medical imaging applications.

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This study is dedicated to the memory of Professor G.E. Giakoumakis, leading member of our team, whose work on phosphor materials has inspired us to continue.

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