QUANTUM NOISE IN X-RAY IMAGE INTENSIFIERS

Tamás Porubszky

Received March 1986, accepted August 1986

ABSTRACT

Explicit expressions are given for the relative variance of the output screen of X-ray image intensifiers, and for detective quantum efficiency as a function of the technical features of the intensifier. The analysis takes into account all conversion stages and physical processes inside the image intensifier and includes a special treatment of scintillation. The results are analyzed numerically both for previously reported approximations and for applying this more recent method of detail visibility calculations.

Keywords: X-ray apparatus, imaging, intensifier

INTRODUCTION

Image noise is one of the important factors of diagnostic image quality. Noise originating from the quantized nature of X-rays is high, especially in intensified radioscopy with low dose rates, and represents a physical limit to the decrease of X-ray intensity.

Noise can be characterized mathematically by the signal-to-noise ratio (S/N) or its reciprocal, relative fluctuation. An image transmission system can be characterized by the decrease of S/N between its output and input, i.e. the detective quantum efficiency (DQE) (Ref 1, pp 192-195) which is defined as

$$DQE = \frac{\left(S/N\right)_{\text{out}}^2}{\left(S/N\right)_{\text{in}}^2}$$
(1)

Noise in X-ray image transmission systems has an extensive literature^{2, 3, 4}, but few papers deal with noise in electron-optical X-ray image intensifiers (XRII). The earlier ones⁵⁻⁹ give no explicit relation between the S/N of an XRII and its technical features. The most recent work in this field is that of Rowlands and Taylor¹⁰ which does give such an expression and also deals with the problems of DQE of XRIIs both theoretically and experimentally, but without justifying some omissions.

The aim of this paper is to give an explicit expression for the S/N of an XRII based on available data and some theoretical considerations; it contains a mathematical model of scintillation and considers the XRII as a system with all its conversion stages. Numerical analysis of the result gives not only an evaluation of earlier approximations and omissions but may also be applied to detail visibility calculations.

Medicor Research and Development Institute (Technical Manager: I. Bíró), H-1389 Budapest, P.O. Box 150, Hungary. Reprints from Dr T. Porubszky.

GENERAL CONSIDERATIONS

An analysis of the multistage image formation of XRIIs must take into account the following noise sources: the incident X-ray beam, transmission of the XRII entrance window and phosphor substrate, X-ray absorption of the input screen (phosphor), K-escape fraction, scintillation of the input screen (number of excitations *and* that of light photons), light collection efficiency, electron emission of the photocathode, electron collection efficiency and scintillation of the output (viewing) screen.

These processes are generally treated as multiplications. To express the fluctuation of the output image (i.e. photon fluctuation of the viewing screen) one has to know the statistical distributions of the incident beam and of the 'multiplication' of the individual stages as well.

The earliest and most widely referenced paper on this subject is that of Sturm and Morgan¹¹, which unfortunately contains several mathematical inaccuracies, as has been pointed out^{12, 13}.

A particle multiplication process can mathematically be written as $n_i = k_i n_{i-1}$ where n_i is the number of particles and k_i is the multiplication factor of the *i*-th stage (see *Figure 1*); the case when $k_i < 1$ is not excluded. The individual stages are independent, which means that variances $\sigma_{k_i}^2$ and $\sigma_{n_{i-1}}^2$ are independent of each other; such processes have been analysed^{6, 14-16}.

The most general expression giving the quadratic relative fluctuation (relative variance) of the output (see Ref 17)

$$\frac{\sigma_{n_m}^2}{\overline{n}_m^2} = \frac{1}{\overline{n}_0} \sum_{j=1}^m \frac{\sigma_{kj}^2}{\overline{k}_j \prod_{i=1}^j \overline{k}_i} + \frac{\sigma_{n_0}^2}{\overline{n}_0^2}$$
(2)

where \bar{n}_0 is the expectation value of the number of the incident particles corresponding to the area of an image point (picture element) and an integration time (in radioscopy, that means principally the



Figure 1 Schematic representation of a particle multiplication process $(n_{i-1}: input number of particles, k_i: multiplication factor, n_i: output number of particles)$

storage time of the human eye) \bar{n}_m is the expected number of output particles corresponding to the same picture element and time interval, \bar{k}_j is the expected value of the *j*-th multiplication factor and σ^2 s are the corresponding variances. Expected values of random variables are denoted by a bar above () in this paper. (*Note:* It can easily be seen that equation 2 is valid for a one-stage process, for n_0 and k_1 having either Poisson or Bernoulli distribution. Its general validity can be proved then by induction).

Definition of the system

The XRII may be considered as a (coherent) system, the input signal is therefore an X-ray beam incident upon the entrance window of the XRII, and the output signal is the visible light (photons) emitted by the output screen. We intend to characterize the DQE of the XRII as a transmission system by taking into account all its physical processes.

The incident beam

In contrast to Rimkus and Baily¹³, but in accordance with all other authors, we start from the Poisson distribution of X-ray photons emitted from the X-ray tube³. On average $\bar{n} = \varphi a \tau$ photons fall onto an area *a* during a time interval τ from an X-ray beam characterized by a fluence rate φ . Because of the Poisson distribution the variance of *n* is $\sigma_n^2 = \bar{n}$, i.e. its standard deviation is $\sigma_n = (\bar{n})^{1/2}$

Absorption and transmission

Absorption has several times been shown to be responsible for effects between the X-ray tube and the output screen of the XRII (X-ray tube window, window of the tube housing, collimator, filter, air, patient, radiographic tabletop, entrance window of the XRII, phosphor substrate, and at last the input screen itself). If exactly n_0 particles fall onto an absorbing layer and for each particle there is a probability p that it will be absorbed (and (1-p) to be transmitted) then the probability of absorption of n_1 particles is given by a Bernoulli distribution. Its expected value is $n_1 = n_0 p$ and its variance $\sigma_{n_1}^2 = n_0 p (1-p)$. If $n_0 = 1$ it follows that for the k₁th 'multiplication factor' of the process:

$$\bar{k}_1 = p;$$
 $\sigma_{k_1}^2 = p(1-p)$ (3)

Changing the meaning of p and (1-p), the same can be said about the distribution of the transmitted particles.

If the distribution of the incident particle beam is Poissonian (therefore $\sigma_{n_0}^2 = \overline{n}_0$) then the distribution of the transmitted (as well as the absorbed) particles remains Poissonian,^{1,3,9,18,19} with an expected value $\overline{n}_1 = \overline{n}_0 p$. The expression of the variance also can be obtained immediately by substituting equation 3 into equation 2. Thus the distribution of the X-ray photons incident upon the entrance window and also of those absorbed by the input screen is Poissonian.

Transmission of the entrance window of the XRII

Let $\overline{n}_0 = \varphi a \tau$ denote the expected value of the number of X-ray photons arriving at an area *a* of the entrance window during a time interval τ , where φ is the fluence rate, furthermore, *t* is the fraction transmitted by both this window and the phosphor substrate, i.e. *t* means their total transmission. Then according to the foregoing, the distribution of the transmitted photons is Poissonian, with an expected value $\overline{n}_1 = \overline{n}_0 t$. In other words: the probability of transmission is *t* for each photon, thus the expected value of the k_1 th 'multiplication factor' is just $k_1 = t$.

The variance also can be obtained indirectly by applying equation 2, substituting $\sigma_{n_0}^2 = \overline{n}_0$,

$$\overline{k_1} = t$$
 and $\sigma_{k_1}^2 = t(1-t)$:
 $\frac{\sigma_{n_1}^2}{\overline{n_1}^2} = \frac{1}{n_0} \left(1 + \frac{1-t}{t} \right) = \frac{1}{\overline{n_0}t}$

This expression, however, for incident particles having a Poissonian distribution, is more easily obtained directly.

Absorption of the input screen. K-escape

If a fraction f of the photons incident upon the input screen is absorbed, it follows that the absorbed photons have a Poissonian distribution with an expected value $\bar{n}_2 = \bar{n}_0 t f$, which may be written as $\bar{k}_2 = f$ (multiplication factor).

K-escape fraction from the phosphor is an important factor in the decrease of S/N ratio¹⁰. Denoting this fraction with $f_{K_{esc}}$, energy of a fraction $f_e = 1 - f_{K_{esc}}$ of the particles takes part in the succeeding processes. Similarly, it can be considered as a 'multiplication' process with a 'multiplication factor' $\bar{k}_3 = f_e$, i.e. the number of particles (still having a Poissonian distribution) which take part in the succeeding processes:

$$\overline{n}_3 = \overline{n}_0 t f f_e \tag{4}$$

Modelling of scintillation

Authors generally treat scintillation as a particle multiplication process for which $\sigma_k^2 = \bar{k}$ where *k* is the multiplication factor^{5, 6, 14, 20}. We wish to give a more precise analysis, which is a modified version of an earlier calculation¹² modified by physical considerations.

Of the X-ray interaction processes only photoeffect has a high probability in the phosphor. An X-ray photon absorbed in a photoeffect produces a photoelectron, the energy of which is expended totally in excitations of electron-hole pairs²¹. The energy needed for one electron-hole pair excitation is equal to about three times the forbidden energy bandgap²². Denoting this energy by E_g and the energy of an X-ray photon with E_1 , one photon produces E_1/E_g excitations. (It can reasonably be assumed that E_1/E_g is an integer). As E_g can be considered a constant, it follows from the conservation of energy that the number of excitations is also strictly determined. In case of n_3 absorbed photons there will be $N_4 = (E_1/E_g)n_3$ excitations. But the number of the emitted light photons will be fewer than N_4 because radiationless transitions are also produced in the phosphor. The difference between absorbed energy and emitted light energy is converted into heat.

The energy emitted in the form of light photons can be obtained with the aid of the energy transformation efficiency (or scintillation efficiency) ε . In case of absorption of one X-ray photon having an energy E_1 the total emitted light energy will be

$$\sum E_2 = E_1 \varepsilon$$

whence the expected number of light photons of energy E_2 is $\sum E_2 / E_2 = (E_1 / E_2)\varepsilon$. In the case of absorption of \overline{n}_3 X-ray photons the expected number of light photons will be $\overline{n}_4 = \overline{n}_3(E_1 / E_2)\varepsilon$. The number of light photons emitted by individual events fluctuates around the mean value $(E_1 / E_2)\varepsilon$.

The probability that a light photon will arise from an excitation, is therefore: $p_1 = \overline{n}_4 / N_4$. For the 'multiplication factor' of the excitation-light photon transition: $\overline{k}_4 = p_1$ and $\sigma_{k_4}^2 = p_1(1-p_1)$. Hence the variance of the number of light photons (in case of \overline{n}_3 absorbed X-ray photons):

$$\sigma_{n_4}^2 = N_4 p_1 (1 - p_1).$$

The multiplication factor of the whole scintillation processes $\bar{k}_5 = \bar{n}_4 / \bar{n}_3$ where \bar{n}_3 is given by equation 4. Hence $\bar{n}_4 = \bar{k}_5 \bar{n}_3$; on the other hand $\bar{n}_4 = p_1 N_4$. Thus the corresponding term to be substituted into equation (2) is:

$$\frac{\sigma_{n_4}^2}{\bar{n}_4^2} - \frac{\sigma_{n_3}^2}{\bar{n}_3^2} = \frac{1}{N_4} \frac{p_1(1-p_1)}{p_1^2} = \frac{1-p_1}{N_4 p_1} = \frac{1-p_1}{\bar{k}_5 \bar{n}_3}.$$
(5)

It follows that $\sigma_{k_5}^2 = \bar{k}_5(1-p_1)$. Those authors who have taken $\sigma_{k_5}^2 = \bar{k}_5$, overestimated the variance (Poisson limit). As we shall see later, this difference causes only a very small (some thousandths) difference in the DQE of an XRII. (The same can be said of scintillation counters). The model which has just been outlined is felt to be physically better founded and it also shows the permissibility of the usual approximations.

The photocathode

The *efficiency of light collection* is very close to 1 even in ordinary scintillation counters. For the input screen of an XRII, consisting of CsI needle crystals which function as light conductors and have practically no self-absorption, it certainly can be taken as 1. Therefore there is no statistical noise increase in this stage.

Photons emitted from the input screen produce *electron emission* from the photocathode. One photon can only remove one electron or none. (As even photons of the smallest energy in the relatively narrow scintillation spectrum can induce photoeffect, this spectrum has no influence on the fluctuation of the number of photoelectrons.) The probability of electron emission is given by the quantum efficiency η . Thus this process can be treated in a similar mathematical manner to absorption. The 'multiplication factor' of the process:

$$\bar{k}_6 = \eta \qquad \sigma_{k_6}^2 = \eta (1 - \eta). \tag{6}$$

As the distribution of the incident light photons, then differ from the Poissonian, the result of substituting equation 6 into equation 2 cannot be reached by simple considerations; since $\bar{n}_5 = \bar{n}_4 \eta$, the corresponding term in equation 2 will be

$$\frac{\sigma_{n_5}^2}{\bar{n}_5^2} - \frac{\sigma_{n_4}^2}{\bar{n}_4^2} = \frac{1 - \eta}{\bar{n}_4 \eta} = \frac{1 - \eta}{\bar{n}_0 t f f_e \bar{k}_5 \eta}$$

The *efficiency of electron collection* onto the output screen, however, is smaller than 1 because of back- scattering. Let ζ denote this efficiency. Repeating the former considerations, the (noise-equivalent) number of electrons absorbed in the output screen can be given by $\bar{n}_6 = \bar{n}_5 \zeta$, furthermore, for the 'multiplication factor' k_7 of the process:

$$\overline{k}_7 = \zeta; \quad \sigma_{k_7}^2 = \zeta(1-\zeta).$$

Thus the coriesponding term in equation 2 will be

$$\frac{\sigma_{n_{6}}^{2}}{\overline{n}_{6}^{2}} - \frac{\sigma_{n_{5}}^{2}}{\overline{n}_{5}^{2}} = \frac{1-\zeta}{\overline{n}_{5}\zeta}$$

The output screen

The expected number of light photons emitted from the output screen will be $\bar{n}_7 = k_8 \bar{n}_6$ where \bar{n}_6 the expected value of absorbed electrons and \bar{k}_8 is the expected value of the corresponding multiplication factor. Therefore

$$\frac{\sigma_{n_7}^2}{\overline{n}_7^2} - \frac{\sigma_{n_6}^2}{\overline{n}_6^2} = \frac{1 - p_2}{\overline{k}_8 \overline{n}_6}$$

XRII NOISE AND DQE

Taking into account the former results, the relative variance of the number of light photons emitted by the output screen of an XRII can be given by the following explicit expression (performing the possible reductions):

$$\frac{\sigma_{n_7}^2}{\bar{n}_7^2} = \frac{1}{\bar{n}_0 t \, f \, f_e} \left(1 + \frac{1 - p_1}{\bar{k}_5} + \frac{1 - \eta}{\bar{k}_5 \eta} + \frac{1 - \zeta}{\bar{k}_5 \eta \zeta} + \frac{1 - p_2}{\bar{k}_5 \eta \zeta \bar{k}_8} \right) = \frac{1}{\bar{n}_0 \times DQE},\tag{7}$$

where equation 1 and $\sigma_{n_0}^2 = \overline{n_0}$ were taken into account. Denoting the parts in brackets by *B*, it can be written that DQE of the XRII is:

$$DQE = t f f_e B^{-1}$$
(8)

In equation 7: $\bar{n}_7 = \bar{n}_0 tff_e \bar{k}_5 \eta \zeta \bar{k}_8$ is the expected number of output photons during a time interval τ corresponding to an area *a* of the input screen,

$$\overline{n}_0 = \varphi a \tau$$

where φ is the fluence rate of the X-ray beam incident upon the entrance window of the XRII,

t the total transmission of the entrance window and the phosphor substrate,

f the fraction of X-ray photons absorbed in the input screen,

 $f_{\boldsymbol{e}}$ the fraction of the absorbed photons remaining after K-escape,

 k_5 and k_8 are the multiplication factors of the input and the output screen,

 p_1 and p_2 are the probabilities of luminous transitions of excitations in the input and the output screen

 η is the quantum efficiency of the photocathode and

 ζ the (noise-equivalent) efficiency of electron collection to the output screen. (Dimensions: $[\varphi] = m^{-2}s^{-1}$, $[a] = m^2$, $[\tau] = s$, all the other quantities in equation 7 are dimensionless.)

Notes.

(i) With direct application of equation 2:

$$\frac{\sigma_{n_7}^2}{\overline{n}_7^2} = \frac{1}{\overline{n}_0} \left(1 + \frac{1-t}{t} + \frac{1-f}{tf} + \frac{1-f_e}{tf f_e} \right) B.$$

It can be seen that this expression equals $[(\overline{n}_0 t f f_e)^{-1} B]$, corresponding therefore to equation 7.

- (ii) As Rowlands and Taylor¹⁰ point out, the value of the DQE depends upon the spatial and the temporal frequency. All investigations carried out so far (including the present work) consider DQE only for very low spatial and temporal frequencies.
- (iii) If the light collection efficiency (from the input screen to the photocathode) is $\xi < 1$, then in *B* of equations (7) and (8) a new term appears as follows:

$$B = 1 + \frac{1 - p_1}{\bar{k}_5} + \frac{1 - \xi}{\bar{k}_5 \xi} + \frac{1 - \eta}{\bar{k}_5 \xi \eta} + \frac{1 - \zeta}{\bar{k}_5 \xi \eta \zeta} + \frac{1 - p_2}{\bar{k}_5 \xi \eta \zeta \bar{k}_8}$$

but when $\xi \approx 1$ its influence remains negligible.

Detail visibility

For an X-ray beam having a given (mean) quantum energy and for given XRII features, taking τ as the storage time of the human eye (radioscopy), the output S/N ratio can be written as

$$S/N = \frac{\overline{n}_{7}}{\sigma_{n_{7}}} = Y(\varphi a)^{\frac{1}{2}}$$
(9)

where Y is a numerical constant. In order to be visible in the presence of fluctuation a given pattern must have a contrast of at least

$$C \approx \frac{k}{S / N},\tag{10}$$

where the value of k lies between 3 and 5 (Refs 4 and 11). In equation 10 C is the image contrast. $(a)^{\frac{1}{2}}$ in equation (9) corresponds to the linear dimension h of the pattern. From equation (9) and (10):

$$C \approx \frac{k}{Yh(\varphi)^{1/2}}.$$
(11)

Equation 11 is a general relation between C, h and φ (for given Y and k) from which one can calculate the minimum φ needed to make a pattern visible with a given h and C, or the smallest visible detail diameter for given φ and C. This limiting resolution, obviously, expresses the limiting role of quantum noise alone. Therefore other properties of the image transmission system must also be considered. (Limiting resolution from contrast transfer — expressed usually with modulation transfer functions — cannot be exceeded.)

Numerical calculation of DQE values

Let us apply the previous results for the so-called III, IV and IVE generation of XRIIs made by Thomson-CSF (France). Most of the data are taken from the work of Driard *et al.*⁸ and other Thomson product guides. The data used and the calculated values are shown in *Table 1*, the following must be added to it:

The radiation of a 7 mm aluminium half value layer for which the X-ray absorption of the input screen is given, corresponds to U = 74.7 kV X-ray tube voltage and x = 22 mm Al total filtration according to Mika and Reiss²³. The calculated mean energy weighted by fluence for these parameters is $\overline{E} = 53.9$ keV (Ref. 24). Thus the approximation $E_1 = 54$ keV for the absorbed photons is reasonable. The excitation energy of electron-hole pairs in a CsI input screen can be estimated as $E_g = 20$ eV (Ref. 22). From these data the number of excitations originating from absorption of an X-ray photon is $N_4/\bar{n}_3 = E_1/E_g = 2700$. On the other hand, as the energy conversion efficiency is $\varepsilon = 0.08$ and energy of an emitted light photon is $E_2 = 1000$.

3 eV (Ref. 8), their number: $\bar{k}_5 = \bar{n}_4 / \bar{n}_3 = (E_1 / E_2)\varepsilon = 1440$. Hence the probability of a photon emitting transition: $p_1 = \bar{n}_4 / N_4 = 1440 / 2700 = 0.533$.

Generation		III	IV	IVE
Source data	t	0.75^{a}	0.75	0.85
	f (54 keV)	0.60	0.75	0.78
	B^{-1}	0.99	0.99	0.99
	DQE (60 keV)	0.34	0.44	0.52
Fitted	$f_{ m e}$	0.91	0.91	0.91
Calculated	d/µm	191	289	315
	DQE (54 keV)	0.41	0.51	0.60
	f(60 keV)	0.50	0.65	0.68

^aestimated value

t total transmission fraction of the entrance window and phosphor substrate

f fraction absorbed in the input screen

 f_e remaining fraction after K-escape

d thickness of the input screen (for 100% density)

DQE detective quantum efficiency

B see equations (7) and (8) of the text

Quantum efficiency of the photocathode can be taken as $\eta = 0.15$ (Ref. 8) while the efficiency of the electron collection is about $\zeta = 0.75$ (Ref. 10). As $\bar{k}_8 \approx 10^3$, the last term in *B* is smaller than the others by some orders of magnitude, there is no significance in the value of p_2 .

From these data one can obtain $B^{-1} = 0.993$, independently of generation (to three digits).

The assumed thicknesses of the input screen d were calculated from

$$f=1-e^{-\frac{\mu}{\rho}\rho d},$$

where μ/ρ is the mass attenuation coefficient and ρ is the density of CsI. The value of μ/ρ was calculated from μ/ρ values of Cs and I by weighting by fraction by weight and making a linear interpolation on a log-log scale between the tabulated values of Plechaty et al.²⁵ The density of the needle crystals was assumed to be 100%, i.e. $\rho = 4.5$ g cm⁻³. Using the thicknesses *d* calculated from the data of absorption at 54 keV, the value of *f* can be calculated for any energy. As values of DQE are usually given for 60 keV, we attempted to complete the data series for these two values (i.e. 54 and 60 keV).

The value of f_e was subsequently chosen in order to be compatible with the other results.

Numerical examples for detail visibility calculations

For U = 74.7 kV constant X-ray tube voltage, 2 mm Al equivalent inherent filtration and 25 cm water phantom (which corresponds to a large patient) the mean X-ray beam energy: $E_1 = 54$ keV. Then for i = 1.5 mA tube current the fluence: $\varphi = 4.8 \ 10^5 \text{ cm}^{-2} \text{ s}^{-1}$ (Ref. 24). In this case quantum noise is clearly visible. Substituting the previous numerical values in equation 7 (for IVE generation):

$$\frac{\sigma_{n_7}}{\bar{n}_7} = 1.29 \frac{1}{(n_0)^{1/2}},\tag{12}$$

i.e. the fluctuation of the output is 1.29 times greater than that of the input beam. (For III and IV generation this factor is 1.56 and 1.40, respectively.)

Taking $\tau = 0.2$ s, from eq. (12):

$$S/N = \frac{\bar{n}_{7}}{\sigma_{n_{7}}} = 0.78 \left(\bar{n}_{0}\right)^{1/2} = 0.78 \left(\pi a \varphi\right)^{1/2} = 0.35 \left(a \varphi\right)^{1/2}.$$
 (13)

Substituting $\varphi = 4.8 \ 10^5 \ \text{cm}^{-2} \ \text{s}^{-1}$ and then substituting equation 13 into 11, for a contrast C = 5% (in this case S/N ratio = 100 is needed), the smallest perceptible size: $h = (a)^{1/2} = 0.42$ cm. Therefore the limiting resolution originating from quantum noise is 1.1 lp/cm.

For a contrast of 10% this would be 2.2 lp/cm.

Increasing the X-ray tube current to i = 6 mA, fluence will be proportional, i.e. $\varphi = 1.9 \ 10^6$ cm⁻² s⁻¹. Then for the same contrast values the limiting resolution will be 2.4 and 4.8 lp/cm, respectively.

CONCLUSIONS

The calculations show how the physical processes taking place in XRII tubes increase the noise and at the same time prove the admissibility of some earlier unjustified omissions and approximations.

It can be seen from equation (8), taking into account that for current XRIIs $B^{-1} \approx 1$, that f_e is determined by such physical processes as can be affected only indirectly (e.g. by increasing the phosphor thickness). The DQE value of XRIIs can, in practice, be increased only by increasing the transmission of the entrance window and the phosphor substrate, and increasing the absorption of the input screen. (The thickness of the input screen is always a compromise between X-ray absorption and light collection efficiency.)

Comparing our results (*Table 1*) with those of Rowlands and Taylor¹⁰, there is generally a good agreement. However, in our opinion, in the theoretical considerations of Rowlands and Taylor the number of light photons emitted from the input screen is overestimated, (their estimation refers rather to the number of excitations) while the number of photoelectrons

emitted by the photo-cathode is underestimated, so that the two differences compensate each other. The only significant difference appears in values of K-escape fraction. Fitting the results to the data given in the Thomson-CSF product guide, was possible only in this manner; the reason of this difference will be examined in future experiments.

Accuracy can be further increased if we take into account - instead of using mean energy - the X-ray spectral distribution and the energy dependence of transmission and absorption. The value of such a correction, however, is probably small.

Detail visibility calculations give the absolute limiting resolution originating from the quantized nature of X-rays. According to our previous work²⁴ the input signal contrast can be calculated for given X-ray tube voltage, and voltage and current waveforms. Multiplying this by the modulation transfer function of the image transmission system, the image contrast can be obtained as a function of the spatial frequency of the object. Taking into account the contrast sensitivity of the human eye, it can be determined whether or not a given tissue difference will be visible with a particular X-ray equipment and tube voltage. If so, and using our method, both the minimum fluence rate (thus also the dose rate) and the corresponding X-ray tube current necessary to ensure a really perceptible difference, can be calculated.

ACKNOWLEDGEMENT

The author is greatly indebted to Dr S. Keszthelyi-Lándori (Gamma Works, Budapest) for the valuable discussions concerning the properties of scintillators.

REFERENCES

- 1 Barrett, H.H. and Swindell, W. Radiological Imaging. The Theory of Image Formation, Detection and Processing. Vol. 1. Academic Press, New York, 1981
- 2 Wagner, R.F. Toward a unified view of radiological imaging systems. Part II: Noisy images. Med. Phys. 1977, 4, 279
- 3 Strid, K.-G. Significance of quantum fluctuations in roentgen imaging. Acta Radial. Oncology, 1980, 19, 129
- 4 Evans, A. L1. The Evaluation of Medical Images. Medical Physics Handbooks 10. Adam Hilger, Bristol 1981.
- 5 Dolon, P.J. and Niklas, W.F. Quantum statistics of X-ray image intensifying systems for fluoroscopy. In: *The Reduction of Patient Dose by Diagnostic Radiologic Instrumentation*. (Eds. Moseley, R.D. Jr. and Rust, J.H.) Charles C. Thomas, Springfield 1964, pp 154-180
- 6 Albrecht, C. Noise sources in image intensifying devices. In *Diagnostic Radiologic Instrumentation. Modulation Transfer Function.* (Eds. Moseley, R.D. Jr. and Rust, J.H.) Charles C. Thomas, Springfield 1965 pp. 291-311
- 7 Reitsma, H. Improved image perceptibility in television fluoroscopy. Medicamundi 1973, 18, 36
- 8 Driard, B., Georges, J.-P. and Guyot, L.-F. Image intensification in medical and industrial radiology. Thomson-CSF NTV 6123 (1976)
- 9 Roehrig, H., Nudelman, S., Fisher, H.D., Frost, M.M. and Capp, P.M. Photoelectronic imaging for radiology. *IEEE Trans. Nucl. Sci.* 1981, NS-28, 190
- 10 Rowlands, J.A. and Taylor, K.W. Absorption and noise in cesium iodide X-ray image intensifiers. Med. Phys. 1983, 10, 786
- 11 Sturm, R.E. and Morgan, R.H. Screen intensification systems and their limitations. Amer.J. Roentgenol. 1949, 62,617
- 12 Porubszky, T. Rontgenkvantumzaj es képerősítő (In Hungarian). Finommech. -Mikrotech. 1978, 17, 267
- 13 Rimkus, D. and Baily, N.A. Quantum noise in detectors. Med. Phys. 1983, 10, 470. (Erratum: 1984, 11, 212)
- 14 Gregg, E.C. Assessment of radiologic imaging. Amer. J. Roentgenol. 1966, 97, 776
- 15 RCA: RCA Photomultiplier Manual, pp. 56-75. RCA 1970.
- 16 Gould, R.G., Judy, Ph.F. and Bjärngard, B.E. Noise-characteristics of a microchannel plate X-ray image intensifier. *Med. Phys.* 1978, **5**, 115
- 17 Harrison, R.M. and Day, M.J. Theoretical and experimental studies of the influence of air kerma rate on threshold contrast in diagnostic fluoroscopy systems. *Phys. Med. Biol.* 1983, **28**, 1319
- 18 Janossy, L. Theory and Practice of the Evaluation of Measurements. Para. 113. Clarendon Press, Oxford 1965
- 19 Fried, D.L. Noise in photoemission current. Applied Optics, 1965, 4, 79
- 20 Keszthelyi, L. Szcintillácios számlálók. Műszaki Könyvkiadó, Budapest, 1964. p. 80 (In Hungarian)
- 21 Birks, J.B. The Theory and Practice of Scintillation Counting Pergamon Press, Oxford 1964, p. 89
- 22 Knoll, G.F. Radiation Detection and Measurement. J. Wiley & Sons, New York 1979, p 256
- 23 Mika, N. and Reiss, K.-H. Tabellen zur Röntgendiagnostik. Teil I. Siemens AG, Erlangen 1973
- 24 Porubszky, T. Calculations concerning voltage ripple of X-ray generators. *Phys. Med. Biol.* 1986, **31**, 371
- 25 Plechaty, E.F., Cullen, D.E. and Howerton, R.J. Tables and Graphs of Photon-Interaction Cross Sections from 0.1 keV to 100 MeV derived from the LLL evaluated-nuclear-data library. Report UCRL-50400, vol. 6, rev. 3, 1981