AN INDEX TO COMPARE THE PERFORMANCE OF SCINTIGRAPHIC IMAGING SYSTEMS

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The contrast-efficiency function is a suitable means of describing the spatial-resolution characteristics of a system but cannot be used to describe total-system performance since it does not include the factor of plane sensitivity. The concept of performance index is presented as a measure of total-system performance. This function includes the system's spatial resolution in terms of the contrast-efficiency function and plane sensitivity as measured by statistical fluctuation. The index has special merit in that it measures how well a given system will reproduce a specific input problem in the image plane. Therefore, when used to compare the performance of various systems or to study the effect of varying a parameter within a given system, all comparisons are made on the basis of how well the systems in question are able to reproduce an object having specific geometry in the image plane for equal times of observation. The results of experimental studies show that for small lesion sizes, systems having high spatial resolution give the best performance whereas for large lesions, the system having the highest plane sensitivity gives the best results. For intermediate lesion sizes, the concept shows, as should be expected, that the system having the optimum trade-off between spatial resolution and plane sensitivity for the lesion size evaluated gives the best performance.

One problem, which is of minor importance, is that the index does not as yet include the effect of a change in MTF with distance off the focal plane. This effect will only be important when evaluating the performance of lesion detection when the size of the lesion is such that the MTF varies significantly over the source distance. Application of the performance index concept in future investigations includes the following: (A) pulse-amplitude discrimination level studies on radioisotopes in clinical use; (B) total-system performance studies of collimators available for scintigraphic imaging systems; and (C) comparison of the systems performance of the various stationary scintigraphic imaging devices currently available.

Performance of a particular collimator or one imaging system relative to another is not completely evaluated by a measure of spatial resolution or a description of the line-spread function alone. Only a few attempts have been made to compute an index which combines the spatial-resolution and sensitivity characteristics for a given set of scanning conditions (*1–4*) to reflect the trade-off between these two parameters. Recently Rollo and Schulz (*5*) presented the contrast-efficiency function as an index of spatial-resolution performance.

This function, when combined with plane-sensitivity measurements, provides a quality factor which bridges the gap between the various formulations currently available. The function is unique in that it includes plane sensitivity and takes into account the modulation-transfer function (MTF) of the scanning system for all spatial-frequency components and characterizes the system on the basis of how this

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function matches the frequency spectrum of the object being scanned. The function has the special advantage of permitting a comparison of performance between systems as well as within a single system where parameters affecting spatial-resolution and plane-sensitivity characteristics are varied.

In this paper the contrast-efficiency concept is reviewed and then the quality factor called the "performance index" is described. Some applications of the performance index are presented.

METHOD

Theory: contrast-efficiency function. The contrast-efficiency function proposed by Rollo and Schulz takes into account the overall MTF of the scanning system as well as the spectrum of the lesion being scanned. Although the concept presented deals with spherical lesions, the approach is equally applicable to lesions having any geometry which can be described mathematically.

Basically, the method involves determining the lesion-spread function $C_L(r)$, i.e., the spatial-distribution function of a spherical lesion from the projected volume distribution of the sphere as a function of its radial distance from the central axis. The normalized lesion-transfer function $L(f)$ of each spherical lesion having a radius r is then obtained from the Fourier transform of that lesion's spread function.

When the lesion is scanned by an imaging system having an overall normalized MTF, MTF(f), the frequency spectrum of the image of the scanned lesion, i.e., the image-transfer function $I(f)$, is the following:

$$I(f) = L(f)\,\mathrm{MTF}(f) \tag{1}$$

The inverse transform of this composite function gives the image-spread function $C_s(r)$, i.e., the one-dimensional spatial-distribution function of the scanned spherical lesion.

Thus

$$C_s(r) = \int_{-\infty}^{\infty} I(f)e^{i2\pi fr}\cdot df \tag{2a}$$

or

$$C_s(r) = 2\pi\int_0^{\infty} fI(f)\,J_0(2\pi fr)df \tag{2b}$$

where f is the spatial frequency, J_0 is the zero-order Bessel function of first kind, r is the radial distance from the central axis of lesion, and $I(f)$ is the image-transfer function of the scanned lesion.

For an actual image, the expected lesion count density profile is given by $A\Lambda\tau C_s(r)$ where A is the total (effective) activity in the lesion, Λ is the imaging system plane sensitivity, and τ is the imaging time. If one assumes an effective background-activity concentration B, then at the origin ($r = 0$) the image contrast of the scanned lesion C_i is obtained:

$$C_i = AC_s(0)/B = 2\pi A\int_0^{\infty} fI(f)df/B \tag{3}$$

It should be emphasized that C_i represents the image contrast of the scanned lesion, having radius r, and represents the contrast one would obtain with an imaging system having the normalized response function MTF(f).

If one were to scan a lesion having a lesion-transfer function $L(f)$ with a system having perfect resolution, i.e., MTF(f) unity at all spatial frequencies, the resulting frequency spectrum of the scanned lesion would be $L(f)$, the spread function of the lesion alone. Consequently, in the presence of the same background activity B assumed in developing Eq. 3, the contrast C_o, obtained when a lesion is scanned with this perfect system, is the following:

$$C_o = AC_L(0)/B = 2\pi A\int_0^{\infty} fL(f)df/B \tag{4}$$

Therefore, C_o represents the object contrast, i.e., the contrast one would obtain by scanning the lesion with an imaging system having perfect resolution. The ratio of the image contrast C_i to the object contrast C_o, is defined to be the contrast efficiency E_c, and is given by the following equation:

$$E_C = C_i/C_o = \int_0^{\infty} fL(f)\,\mathrm{MTF}(f)df\Big/\int_0^{\infty} fL(f)df \tag{5}$$

It can be seen that this function is somewhat analogous to the modulation-transfer function in that it is a measure of the efficiency of transferring the object contrast to an image contrast. The contrast-efficiency function differs from the MTF in that the former measures the efficiency of transferring the contrast of an object having a mathematically definable geometry (sphere) into an image contrast as a function of sphere radius whereas the latter measures the efficiency of the system in transferring the contrast corresponding to each sinusoidal frequency comprising the object to the recorded image.

In practice, the contrast-efficiency values are determined in the following way. The MTF(f) of the system to be evaluated is determined as a function of spatial frequency from experimentally measured line-spread function data. The lesion-transfer functions are then computed for each lesion size as a function of spatial frequency. For each lesion size, Eq. 5 is then used to determine the corresponding contrast-efficiency values as a function of lesion radius. For ease of computation, a computer program is utilized that requires only the line-spread function data as input.

The contrast-efficiency function provides a quantitative measure of how well the transfer function of a given imaging system matches the spectrum of the lesion being scanned. As such, it permits comparison of resolution performance between systems as well as within a single system where parameters affecting resolution characteristics are varied.

Theory: performance index. The contrast-efficiency function as defined thus far quantitatively measures only the spatial-resolution characteristics of the imaging systems being evaluated. In order to compare the performance of imaging systems fully the defining function should also include a measure of sensitivity, S. One very useful measure of sensitivity is the statistical fluctuation ϵ, which is defined as

$$\epsilon \propto 1/\sqrt{S} \qquad (6)$$

where S is plane sensitivity in cycles per second per microcurie per square centimeter for a moving detector and cycles per second per square centimeter per microcurie per square centimeter for a stationary device.

It is proposed in this paper that a performance index be defined which is the product of the contrast efficiency computed for a given lesion having radius r, $E_c(r)$, and the reciprocal of the statistical fluctuation of the system ϵ, both measured for a given set of experimental conditions. This equation is as follows:

$$\Phi = E_c(r) \cdot 1/\epsilon \qquad (7)$$

If one now substitutes the appropriate equations for each variable from Eqs. 5 and 6

$$\Phi = \left[\int_0^\infty fL(f)\, MTF(f) df \Big/ \int_0^\infty fL(f) df \cdot \right] \sqrt{S} \qquad (8)$$

The resulting function Φ is a performance index which includes the effects of the imaging system's spatial resolution and plane sensitivity. The function is particularly attractive in that it can be used to compare the performance of various imaging devices toward detecting any mathematically describable distribution function. It would appear, however, that the most interesting case is that of comparing the detection of spherical lesions.

Basically, the performance index is a measure of how well a spherical object of radius r will be reproduced in the image plane by the system having the experimentally measured spatial-resolution characteristics and statistical fluctuation used in the calculation. Obviously, the higher the index, the better the resulting image.

The index can be used to compare the performance of different systems as well as to study the effect of varying a parameter within a single system. In either case, it is important to point out that in all comparisons the index is a measure of the relative detectability of spherical lesions for equal data accumulation times. Examples demonstrating the applications of the contrast efficiency and performance index will now be given.

TABLE 1. OHIO-NUCLEAR COLLIMATORS

	5C	5D	5E
Geometric focal length (cm)	8.8	8.8	8.7
Thickness of collimator (cm)	8.7	8.4	6.7
Diameter of hole at crystal (cm)	0.62	0.89	0.95
Number of holes	169	91	61
Radius of resolution (cm)	0.63	0.93	1.23
Geometric efficiency (cm^2)	0.884	0.21	0.35

RESULTS

Two experiments were performed. The first experiment compares the performance of a single rectilinear scanning system when three different collimators are used whereas the second involves the evaluation of pulse-amplitude discriminator settings on a rectilinear scanning system.

The Ohio-Nuclear Model 54 Dual 5 scanner was used in a study to compare the performance of this system with the 5C, 5D, and 5E collimators. The geometric data associated with each collimator is shown in Table 1.

The radioisotope ^{99m}Tc was used with a pulse-amplitude discriminator setting of 130–170 keV. For each collimator, line-spread function measurements were made in 1-mm steps with a 32-cm-long thin plastic (0.5 mm i.d.) line source suspended in a water tank with the face of the collimator at the water surface. In each case, 7.62 cm of water existed between the source and the face of the collimator and 15 cm of water was present below the source. Plane-sensitivity measurements were made (cps/μCi/cm^2) by placing a Lucite sheet phantom (20 $\times$ 20 $\times$ 0.5 cm) filled with ^{99m}Tc having a known activity at a depth of 7.62 cm in the water tank. The associated modulation-transfer functions and contrast-efficiency functions were calculated for each collimator using the previously described technique. The contrast-efficiency values for each lesion size were then multiplied by the square root of the plane sensitivity to determine the performance index as a function of lesion radius for each collimator. The modulation-transfer function, contrast efficiency, and performance-index curves are shown in Figs. 1, 2, and 3, respectively.

In Fig. 1 it can be seen that all the systems have equivalent spatial-resolution performance at spatial frequencies below 0.12 cycles/cm. In the frequency range between 0.12 cycles/cm and 1.5 cycles/cm, the best spatial resolution is achieved by the 5C collimator system, Curve A, followed in order by the 5D system, Curve B, and the 5E system, Curve C. Above 1.5 cycles/cm the high-frequency oscillations make interpretation difficult. These high-frequency oscillations are an inherent difficulty with MTF curves which may be real consequences of actual spread-function shape under certain conditions but more frequently are related to truncation of spread-function data. In addition, as emphasized in this paper, the MTF approach does not provide a direct measure of how the system being evaluated will change the frequency spectrum of scanned objects of various sizes.

Figure 2 shows that the contrast-efficiency concept compares systems on the basis of how their respective MTFs modify the transfer function of spherical lesions of various sizes, i.e., all systems are compared on the basis of how well they perform on a specific object. It can be seen that for spherical lesions having radii less than approximately 28 mm, the best spatial-resolution performance is obtained with the 5C collimator, Curve A, followed by the 5D collimator system, Curve B, and the 5E system, Curve C. If one compares the relative contrast-efficiency value of each collimated system for lesions having radii greater than 28 mm (Fig. 2) it can be seen that all systems have essentially the same resolution performance, i.e., the contrast efficiency for all systems will be unity. This of course is to be expected since as lesions get bigger, the measured contrast correspondingly increases until a value of unity is achieved. The lesion size at which unity contrast is obtained should of course depend upon the spatial-resolution characteristics of the system utilized. This is also exemplified in Fig. 2 where it can be seen that the ultimate contrast is obtained for lesion radii of 20 mm, 28 mm, and 34 mm for the 5C, 5D, and 5E collimated systems, respectively.

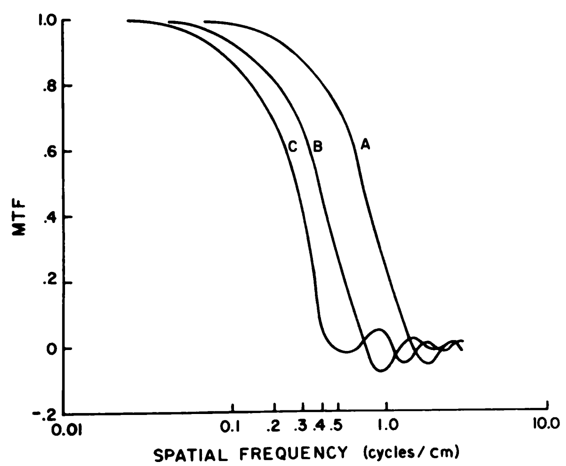

FIG. 1. Normalized modulation-transfer function (MTF) versus spatial frequency for Ohio-Nuclear Model 54 scanner. Curve A is for 5C collimator, Curve B for 5D collimator, and Curve C for the 5E collimator. A ^{99m}Tc line source placed at depth of 7.62 cm within water tank was used in all cases to determine line-spread function data. Pulse-height selector was set at 130–170 keV.

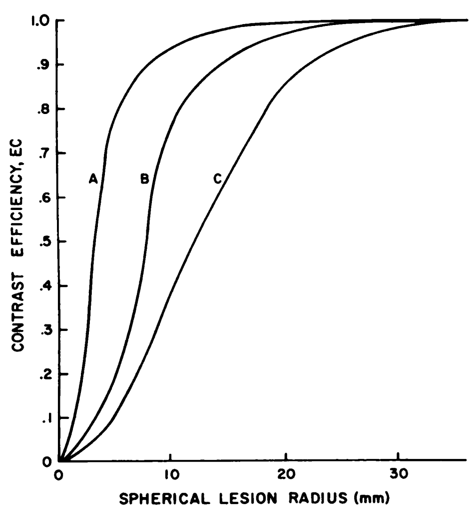

FIG. 2. Contrast efficiency as function of spherical lesion radius for Ohio-Nuclear Model 54 scanner. Curve A is for 5C collimator, Curve B is for 5D collimator, and Curve C for 5E collimator. All curves are for ^{99m}Tc source in 7.62 cm of water with pulse-height selector set at 130–170 keV.

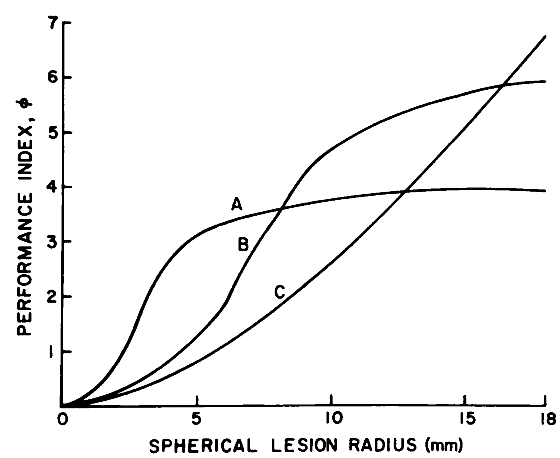

FIG. 3. Performance index Φ as function of spherical lesion radius. In all cases Ohio-Nuclear Model 54 scanner was used with pulse-height selector setting of 130–170 keV. Source measurements were made at depth of 7.62 cm in water-scattering medium. Curve A is 5C collimator; Curve B, 5D collimator; and Curve C, 5E collimator.

Figure 3 shows that when the factor of statistical fluctuation is included in evaluating system performance, the 5C collimator offers the best performance for lesions having radii less than 8 mm whereas the 5D collimator provides a better performance index for lesions having radii between 8 mm and 16 mm. The 5E collimator performs best at radii greater than 16 mm. This interesting result of course reflects the fact that for very small lesions one needs fine spatial resolution for best performance, but as lesion size increases, statistical fluctuation considerations become increasingly more important. For very large lesions sensitivity becomes the predominant factor. This example clearly demonstrates that the trade-off expected between sensitivity and spatial resolution is a function of lesion size.

A second example of the use of the performance index concept involves the use of the Ohio-Nuclear Model 54 Dual 5 scanner and the 5D collimator. In this case, the effect of pulse-height selection was studied for the radioisotope ^{99m}Tc. Line-source measurements were made in water as described in the first experiment for window settings of (A) 95–170 keV, (B) 105–170 keV, (C) 115–170 keV, (D) 125–170 keV, (E) 130–170 keV, and (F) 140–170 keV. Plane-sensitivity measurements were also made on the sheet phantom filled with a known activity of ^{99m}Tc at a depth of 7.62 in water for the same set of window settings.

The line-spread function data were used to determine the modulation transfer-function and contrast-efficiency functions for each window setting. The resulting curves are shown in Figs. 4 and 5. The plane-sensitivity measurements were used to determine the statistical fluctuation values corresponding to each window setting and the results combined with associated contrast-efficiency function data to establish performance index curves for each window setting as a function of lesion radius. The curves are shown in Fig. 6.

Both Figs. 4 and 5 clearly demonstrate that as the baseline window is advanced toward the photopeak, spatial resolution improves. It should again be noted, however, that the high-frequency oscillations present

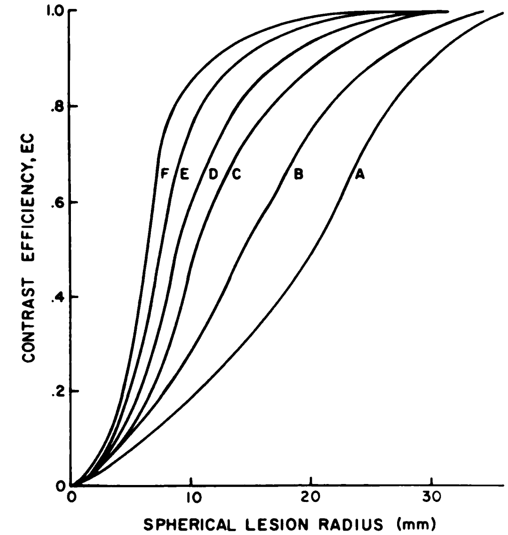

FIG. 5. Contrast efficiency versus spherical lesion radius for various window settings. Model 54 Ohio-Nuclear Scanner was used with 5D collimator. All source measurements were made on ^{99m}Tc at depth of 7.62 cm in water-scattering medium. Window setting for Curve A is 95–170 keV; Curve B, 105–170 keV; Curve C, 115–170 keV; Curve D, 125–170 keV; Curve E, 130–170 keV; and Curve F, 140–170 keV.

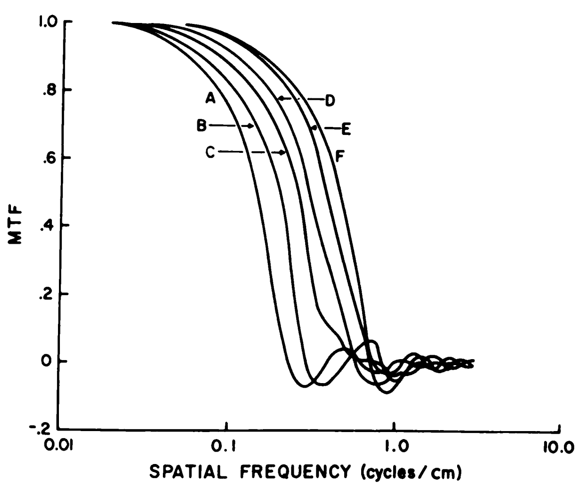

FIG. 4. Modulation-transfer function versus spatial frequency for various window settings. Model 54 Ohio-Nuclear Scanner was used with 5D collimator. All source measurements were made on ^{99m}Tc at depth of 7.62 cm in water-scattering medium. Window setting for Curve A is 95–170 keV; Curve B, 105–170 keV; Curve C, 115–170 keV; Curve D, 125–170 keV; Curve E, 130–170 keV; and Curve F, 140–170 keV.

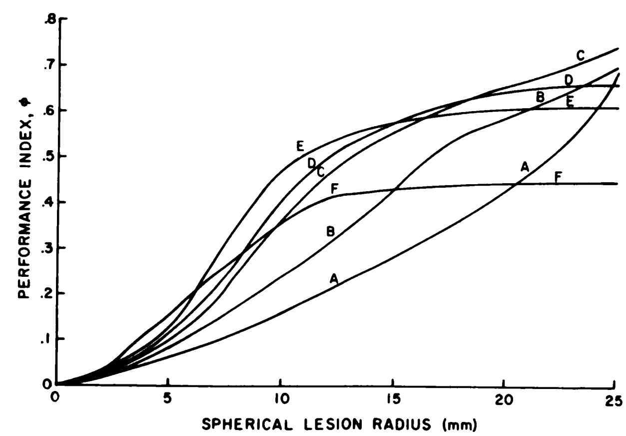

FIG. 6. Performance index, Φ, as function of spherical lesion radius for various window settings. Model 54 Ohio-Nuclear Scanner was used with 5D collimator. All source measurements were made on ^{99m}Tc at depth of 7.62 cm in water-scattering medium. Window setting for Curve A is 95–170 keV; Curve B, 105–170 keV; Curve C, 115–170 keV; Curve D, 125–170 keV; Curve E, 130–170 keV; and Curve F, 140–170 keV. (Φ has been multiplied by 10^{-1}.)

in Fig. 4 make interpretation difficult in the spatial-frequency range over which they occur.

Figure 6 again illustrates, as did Fig. 3, that system performance is not only dependent upon spatial resolution and statistical fluctuation but also the nature of the specific input problem, i.e., the size of the lesion being evaluated. It can be seen from this set of curves that for very small lesions (radii less than 6 mm), the best performance is attained by the 140–170-keV window which of course provides the best spatial resolution whereas the best performance for lesion sizes of clinical interest (radii between 6 mm and 15 mm) is attained by a window setting of 130–170 keV. This setting has also been found to be ideal by Gottschalk and Beck (*1*), using statistical criteria, and by Rollo and Schulz (*6*) using a scintigraphic scan simulation technique.

For lesions having radii greater than 15 mm it is found that lower baseline settings give better performance than higher baseline settings because of improved statistical fluctuation in the presence of adequate spatial resolution. In the two examples presented, the trade-off between spatial resolution and sensitivity is clearly shown to be a function of the size of the lesion being detected. It is therefore evident that the performance of various systems can only be adequately compared when lesion size is taken into account. Because the basis of the performance index concept is the inclusion of the input problem, it is considered to have special value toward evaluating and comparing scintigraphic imaging-system performance.

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