Collimator Technology and Advancements

Robert J. Wilson

University of Tennessee, Memphis, Tennessee

This is the final article in a four-part continuing education series on scintillation camera technology. After studying this article the reader should be able to: 1) understand the differences in collimator design; 2) evaluate the proper utilization of collimators; and 3) discuss the resolution and sensitivity trade-offs in collimator selection.

The purpose of the collimator in nuclear medicine is to limit the field of view of imaging to obtain spatial information concerning the radioactivity distribution. To accomplish this task, an essential trade-off is made between the spatial resolution and sensitivity. Many factors must be considered in this trade-off, including:

1. Resolution.
2. Resolution at depth in patient from the surface of the collimator.
3. Field of view of the image.
4. Septal thickness between the holes of the collimator and penetration of this septa by gamma rays.

A perspective and a description of the single-hole collimator on probes and then a multi-hole focused collimator used for imaging is presented. The properties of collimators and the various camera collimators end the discussion. Various collimators are described in this paper; however, the major discussion centers upon the collimators used with scintillation cameras.

SINGLE-HOLE COLLIMATORS

The single-hole collimator limits the field of view of a scintillation probe to a single organ. The sensitivity for counting is approximately equal over the entire field of view. The most common use of this collimator is for iodine uptake measurements of the thyroid. An example of this "flat-field collimator" is shown (Fig. 1). The sensitivity profile at the working distance illustrates the relatively constant sensitivity over the field of view. The working distance, D, from the scintillation detector is a compromise. The working distance must be sufficiently large so that variations in organ depth (using the inverse square law) do not significantly affect the count measurement, while maintaining enough sensitivity for detection (also determined by the inverse square law). A complete treatment of these trade-offs has been described (*1*). For most probes, the working distance will be 20–30 cm depending on the conditions found in the application, such as crystal radius, collimator length, desired field size, etc.

Similar probes are used in renography and other studies where the retention in the organ of interest is measured, excluding other organs. Two such collimators have been used in a nuclear ventricular probe (*2*), where one collimator defines the ventricle (excluding other areas of the heart) and a second collimator defines a background area of interest.

RECTILINEAR SCANNER COLLIMATORS

A rectilinear scanner produces an image of the radioactivity in a patient by moving the field of view defined by the collimator in a raster motion over the entire region to be imaged. The spatial resolution of the field of view is affected by the geometrical size of the holes in the collimator, penetration of the septa between the holes by the photons, and the ability to reject photons scattered from outside into the geometrically defined field of view. The raster scanning sets up the entire region of the image while the pulse-height analyzer determines the scatter rejection. The purpose of the collimator is a strictly geometrical one, to "look at" a small region of the activity in order to obtain the spatial distribution.

The collimator used for scanners is the multi-hole focused collimator shown in figure 2. The best resolution and best sensitivity are at the focal point, the point where all hole fields of view intersect. Although the counts recorded from an organ remain relatively constant, regardless of its distance from the focal point, information from above and below the focal zone are blurred. Thus, the focused collimator on a rectilinear scanner gives rise to a naturally tomographic image. In other words, in order for something to be imaged with the best resolution, it must be in the focal zone (*3*).

The septa between the holes is a gamma absorbing material, generally lead, of sufficient thickness to maintain the geometric resolution of the hole. Factors affecting septal penetration and its effect on the image are discussed in the next section.

CAMERA COLLIMATORS

In contrast to a rectilinear scanner, where the collimator concentrates on a focal point while the field of view is defined by the raster scan, the collimator for a camera must also determine the entire field of view for imaging. Therefore, when considering collimators for a camera, one must evaluate all five of the parameters describing the collimator. They are:

For reprints contact: Robert J. Wilson, PhD, Radiology Dept., University of Tennessee, 800 Madison Ave., Memphis, TN 38163.

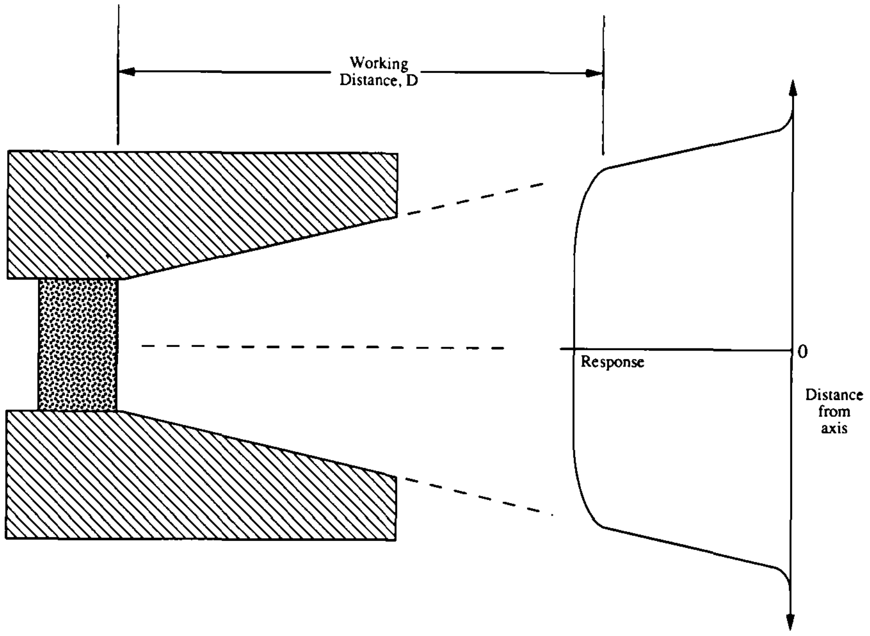

FIG. 1. Single-hole Collimator used on a scintillation probe. The response curve (right) diagrams the relative spatial sensitivity.

1. Resolution.
2. Sensitivity.
3. Resolution at depth.
4. Field of view of the image.
5. Septal thickness and penetration.

Resolution

The most specific measure of resolution is the modulation transfer function (MTF), derived from a measurement of the line spread function (LSF), schematically shown in figure 3. When performing these measurements with a scintillation camera, the overall resolution as measured in the image is determined not only by the resolution of the collimator but also by the intrinsic resolution of the camera. The resolution of a collimator can then be determined from the overall resolution in the image by extracting the intrinsic resolution of the camera (the measured MTF at any spatial frequency being the product of the intrinsic camera MTF and the collimator MTF). Practically, an approximation to the resolution can be measured with a line-pair phantom, consisting of equally spaced lead bars and interspaces. An approximation of the full width at half maximum (FWHM) of the line spread function is:

$$\text{bar width} \approx 1.75\ \text{FWHM}.$$

The intrinsic resolution of the camera must be taken into account to determine the collimator resolution. For the bars, the relationship $R_m^2 = \sqrt{(R_i^2 + R_c^2)}$ can be used to extract R_c, where R_m = measured resolution, R_i = intrinsic camera resolution, and R_c = collimator resolution. A continuously varying line-pair phantom can be used to measure R_m and R_i. An example of such a measurement is shown in figure 4. Note the linear increase of R_c with respect to the distance from the face of the collimator. For the best resolution, the patient must be positioned close to the collimator.

The most commonly used collimators for the scintillation camera are made up of straight bore holes. For purposes of collimator comparison, the geometrical resolution and sensi-

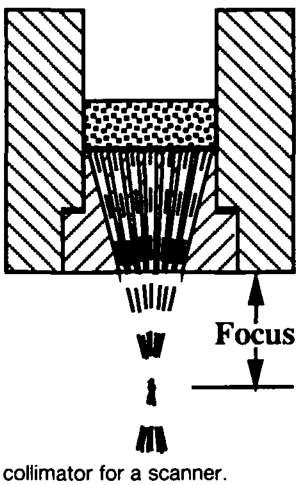

FIG. 2. Focused collimator for a scanner.

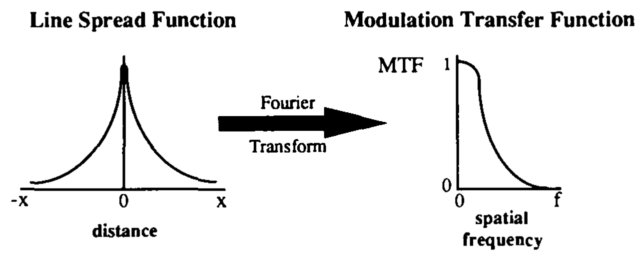

FIG. 3. Schematic rendering of the line spread function and of the modulation transfer function as a measurement of resolution.

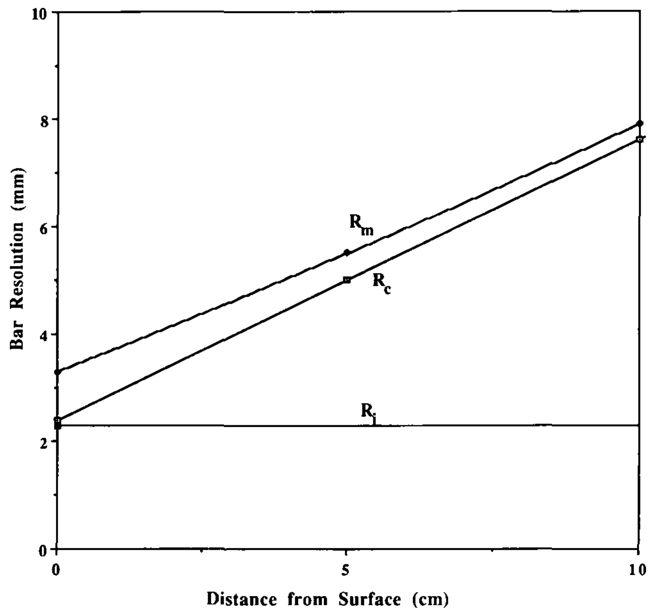

FIG. 4. Measurement of the resolution as the bar width of a line-pair phantom.

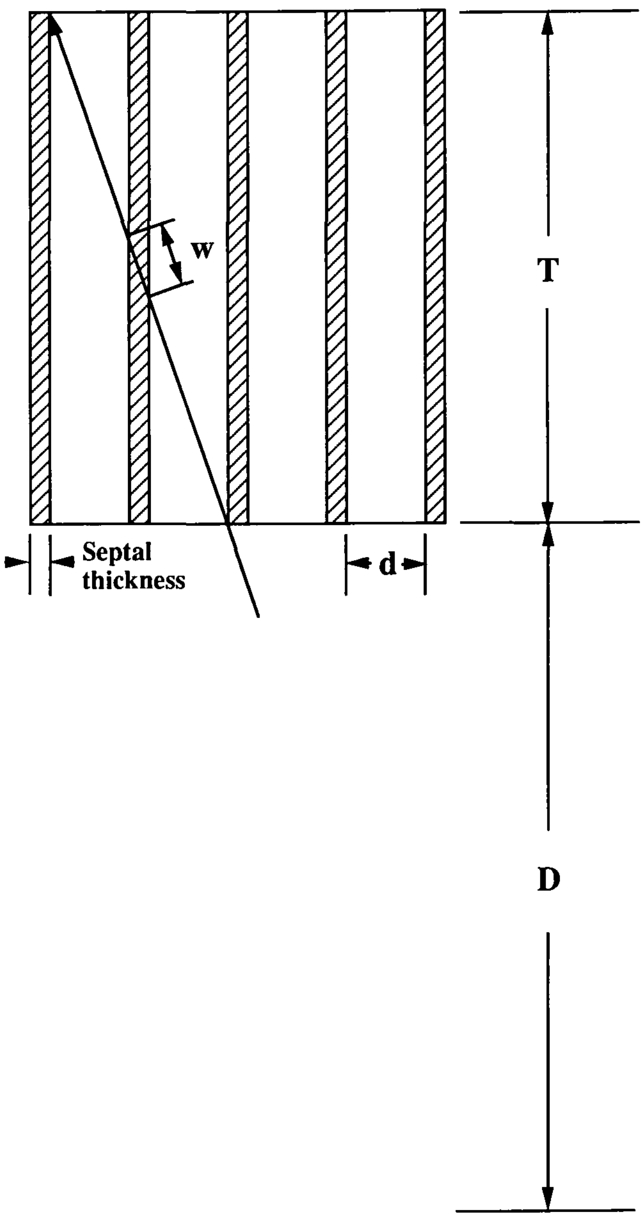

FIG. 5. Straight bore holes and the distances effecting resolution. T is collimator, D is source to collimator distance, d is the whole diameter, and w is the minimum septum path length.

tivity can be derived. The resolution of a single straight bore hole as shown in (Fig. 5) is approximately:

$$\text{FWHM} \approx d(T + D)/T = d(1 + D/T), \qquad \text{Eq. 1}$$

where d = width of the straight bore hole, T = the thickness of the hole, and D = the distance from the surface of the collimator. As measured and shown in figure 4, the resolution FWHM increases linearly with the distance D from the face of the collimator. This relationship can also be used when drawing the straight line between measured values of R_c, which might have measurement error.

Sensitivity

In imaging, uniform sensitivity over the collimator field of view must be maintained for large objects. Since this sheet source sensitivity is independent of the distance from the face of the hole (*3*), the sensitivity can be determined by the efficiency at the surface of the collimator as calculated by using the inverse square law:

$$\text{Efficiency} \approx d^2/16T^2, \qquad \text{Eq. 2}$$

where d equals the diameter of the hole and T is the hole length. Equations 1 and 2 demonstrate that when using straight bore-hole collimators, the following three factors must be considered:

1. The most essential consideration is the trade-off between resolution and sensitivity. A significant problem that cannot be overcome is the dependence of the sensitivity on the square of the resolution. For a fixed thickness collimator, a factor of a two-fold decrease in resolution gives rise to a factor of four-fold decreases in sensitivity. This square dependence precludes the use of a collimator with arbitrarily small resolution.
2. Variation of resolution with depth is mediated by the thickness of the collimator and can never be better than at the surface of the collimator. Therefore, all imaging with scintillation cameras is to be done with the patient next to the collimator whenever possible. This variation of resolution with depth presents a problem for the acquisition of data for single-photon emission computerized tomography (SPECT). In SPECT, if the acquisition of the data requires the camera head to be a considerable distance from the organ of interest, then the resolution in the image is degraded significantly.
3. According to these equations, if resolution is maintained at depth rather than at the surface, then the use of longer and wider holes is appropriate. The trade-off of resolu-

tion and sensitivity is partially mediated by the larger resolution at the surface and a more shallow increase of resolution with distance from the surface of the collimator.

Figure 6 shows the collimators commonly used with the scintillation camera. Each one is discussed in terms of the previously described parameters.

Parallel-Hole Collimators. The most commonly used collimator with the scintillation camera is the parallel-hole collimator. The resolution and sensitivity are given in Equations 1 and 2. The entire field of view of imaging is limited by the area of the crystal, and the image does not have any distortion because the holes are parallel to one another. Thus, each image maintains the proper relative dimensions of the object. The holes are created by either a cast method with septa of lead, or the septa can be formed with lead foil. The advantages of foil construction are the possibility of thinner septa and uniformity of septal thickness, while cast construction can give more rigid construction and thicker septa. Either construction must have rigorous manufacturing controls established to yield consistent performance, even to the variation of the angulation of holes of collimators used for SPECT. The holes may be constructed of various geometrical forms, including circles, triangles, squares, and hexagons. Circular holes give the most efficient geometric hexagonal pattern, but result in a nonuniform septal thickness. Of the various geometrical shapes just described, the hexagonal holes provide the most efficient use of the septa along with a good circular symmetry of the resolution.

The overall sensitivity for organ imaging is determined by the field of view of the camera. A more efficient use of the entire field of view for imaging, e.g., the use of a rectangular field of view camera for whole-body imaging, provides an overall increase in sensitivity for imaging the area of interest. This increase in sensitivity due to more efficient use of the field of view allows more latitude in the questions of sensitivity versus resolution in selecting a collimator.

Terms such as high resolution, high sensitivity, and general purpose are associated with these trade-offs for parallel-hole collimators. The choice of a high sensitivity collimator for fast dynamic studies is obvious, but the interplay of sensitivity and resolution with depth must always be appreciated for the selection of high resolution or general purpose collimators.

Finally, because parallel-hole collimators give rise to little distortion, sizing of organs can be done with parallel-hole collimators by calibration of the imaging device.

Parallel Slant-Hole Collimators. A variation on the paral-

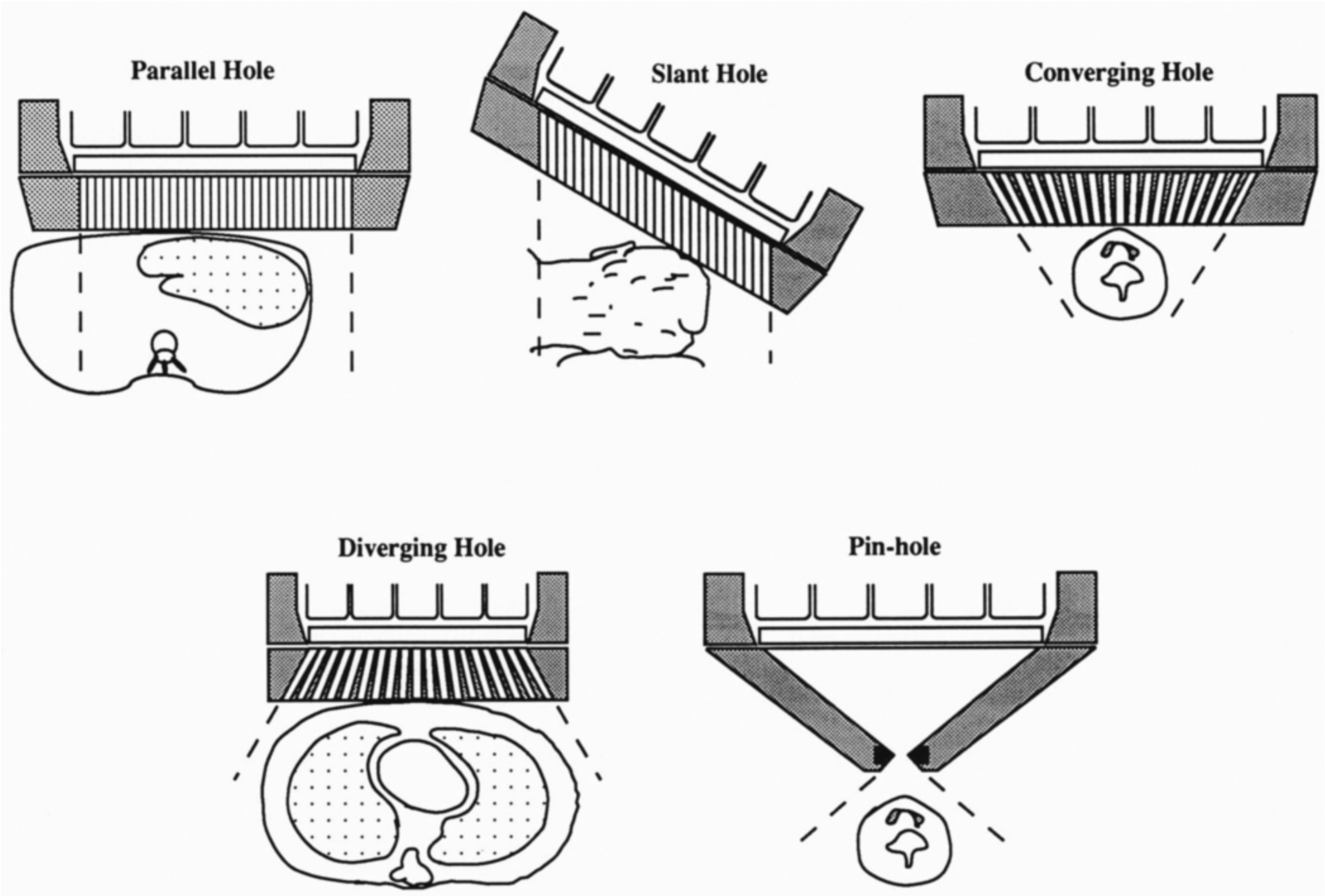

FIG. 6. Commonly used camera collimators.

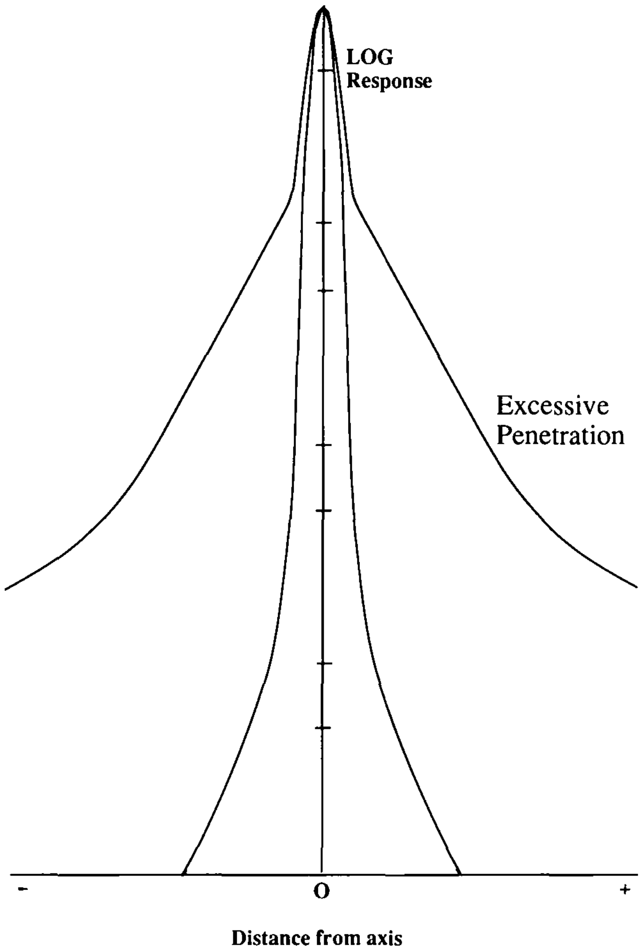

FIG. 7. Line spread fuctions showing effect of septal penetration.

lel-hole collimators, parallel slant-hole collimators are designed so that the parallel holes are at an angle to the collimator face. One of the first uses of these collimators was limited angle tomography, but today they are effective in positioning the collimator face closer to the organ whenever either the body contour or a body part interferes with the camera/collimator housing or gantry. Specifically, in SPECT imaging of the brain, the slant parallel-hole collimator can be positioned close to the head while avoiding the shoulders. A slight increase in sensitivity results from tilting the camera head, with a subsequent linear distortion in the dimension that was tilted. Because more of the camera face views the organ along the tilted dimension, the sensitivity increases, but that dimension is also elongated in the image.

Converging Collimators. In a converging collimator, the holes are in a conical array with the centerlines of the holes intersecting at a point outside the collimator (Fig. 6). A small organ area will be viewed by more holes with a subsequent increase in sensitivity. In comparison with a parallel-hole collimator, the conical form of the holes of a converging collimator gives rise to a magnification. This results in a slight decrease in resolution at the surface but a smaller increase of resolution with increased distance from the collimator. The resolution with distance still increases by virtue of the fact that the resolution of any straight bore hole increases with distance from the face of the collimator. Converging collimators are useful in situations where the field of view may be limited in order to improve the sensitivity by using the entire area of the crystal, for example, in SPECT imaging of the brain. Also, the decrease in resolution with an increase in sensitivity is useful in those situations that do not demand the large fields of view required for large organ imaging, for example, in thyroid imaging. Not only does the sensitivity vary with distance from the collimator, but the dimensions as recorded in the image vary as well. This distortion prevents easy sizing of organ dimensions, limiting the use of converging collimators to imaging of small organs.

To improve the sensitivity while maintaining the resolution in SPECT imaging, cone-beam geometry with a converging collimator has been used (*4*). This is only possible for organs significantly smaller than the field of view of the camera (e.g., the brain). A collimator for SPECT brain imaging converging to two lines of focus, called astigmatic, has been implemented to maximize the sensitivity over a specific volume of the brain (*5*).

Diverging Collimators. Diverging collimators (Fig. 6) have holes in conical array with the center lines of the holes intersecting at a focal point behind the collimator. The holes diverge to increase the field of view beyond the crystal size. At the image edge there is a mild degradation of linearity, uniformity, and resolution. When the field of view of the camera system is not sufficient to visualize the entire area of interest (e.g., in large organ imaging with small field of view cameras), a diverging collimator can be used. With this collimator, sensitivity is sacrificed in order to increase the field of view. In addition, the resolution increase with distance from the collimator has an additional component because of the divergence of the holes and a dimunition of the dimensions of the organ on the crystal. Thus, the larger organ is imaged onto the smaller area of crystal. For many large field of view cameras the diverging collimator is not necessary, but for many small field of view cameras, particularly mobile cameras, diverging collimators still must be used in order to image larger organs, such as the lung. Image distortion results from the diverging collimator because the organ dimensions are smaller on the crystal and dependent on the distance from the collimator face.

A variant of the diverging collimator can be found in some single-pass whole-body imagers. The holes diverge only in the direction across the body in order to encompass the width of the body in the field of view in the camera. This hybrid system maintains the individual hole resolution along the body, but the resolution across the body is increased additionally by the divergence of the holes.

Both the diverging and converging collimators give a distortion of the image; the size of the image recorded is dependent upon the distance from the surface of the collimator. Subsequently, the sensitivity varies with the distance.

Pinhole Collimators. The pinhole collimator magnifies the image of the object on the detector face by allowing the gamma rays to travel in straight lines through an aperture in lead. The geometrical resolution is given by:

$$FWHM = d(A + D)/A, \qquad \text{Eq. 3}$$

where A = the distance of the aperture to the detector, D = the distance to the object, and d = the aperture width. The sensitivity is given by:

$$\text{Efficiency} = d^2/16D^2$$

As illustrated, the resolution for small organs can be selected almost as small as desired, but the sensitivity is simply that of a single hole. The overall sensitivity of the converging collimator, with its many holes, is significantly larger, but the resolution may not be satisfactory. Pinhole collimators suffer from the problem that they also have distortion in the image, which means that they are only appropriate for relatively thin body sections.

An added benefit of pinhole collimators is that they can be used at higher energies without any significant septa penetration. In contrast, an entirely different multi-hole collimator with larger septa must be used at high energies to avoid excessive septal penetration.

Septal Penetration

Septal penetration depends upon the energy of the gammas emitted by the source (not just those that are imaged). High-energy sources and sources which have high-energy contaminants or contributions play an important role in the selection of collimators. It is obvious that the septa play a role in the overall sensitivity of a collimator since there is less sensitivity if septa cover the crystal.

Figure 7, which illustrates effective septal penetration, shows the LSF for two collimators, one with sufficient septal thickness and one with septa too thin, and hence with septal penetration. The tails of the line spread function contribute to an overall "background activity," while the widening of the FWHM can be best appreciated as an increase in the resolution. The amount of permissible septal penetration is dictated by the acceptable resolution loss of the image. Satisfactory images should be obtained if the attenuation along the minimum path length, w, (see Fig. 5), is at least 95% (*6*). After W is determined, the septal thickness can be calculated.

Except for extremely thin septa, the 167 keV gamma ray from ^{201}Tl and the 159 keV gamma ray from pure ^{123}I generally present no problem for the low energy (^{99m}Tc) collimators. The greater energies of the gamma rays from ^{111}In, ^{67}Ga, and impurities in ^{123}I demand greater septal thickness for acceptable septal penetration.

SUMMARY

The essential trade-off of resolution versus sensitivity must be made in the choice of a collimator for imaging. An appreciation of the various factors affecting this choice has been given. Once the selection of a number of collimators has been made, the following components have to be appreciated on each use: (1) Resolution, (2) Sensitivity, (3) Resolution at depth, (4) Field of view, and (5) Septal thickness.

REFERENCES

1. Hine GJ, William JB. Thyroid radioiodine uptake measurements. In: Hine GJ ed. *Instrumentation in Nuclear Medicine, Vol. 1*. New York: Academic Press 1967:327–350.

2. DePuey E, Mathur V, Werp P. Clinical standardization of the new dual cardiac probe (Abstract). *J Nucl Med* 1978:19:731.

3. Harris CC. Collimator detectors. S.N.M. Audiovisual NM9.

4. Jaszczak RJ, Floyd CE, Manglos SH, et al. Cone-beam SPECT: Experimental validation using a conventionally-designed converging collimator (Abstract). *J Nucl Med* 1986:27:930.

5. Hawman EG, Hsien J. An astigmatic collimator for high sensitivity SPECT of the brain (Abstact). *J Nucl Med* 1986:27:930.

6. Anger HO. Radioisotope camera In: Hine, GJ ed. *Instrumentation in Nuclear Medicine, Vol. 1*. New York: Academic Press, 1967.