OPTIMUM DIMENSIONS OF PARALLEL-HOLE, MULTI-APERTURE COLLIMATORS FOR GAMMA-RAY CAMERAS

Eldon L. Keller

Westinghouse Research Laboratories, Pittsburgh, Pennsylvania

The recent introduction of stationary gamma-ray cameras in several forms (*1–5*) has provided increased speed for the *in vivo* determination of radioisotope distributions in many of the human organs. In most cases, these instruments are equipped with parallel-hole, multi-aperture collimators which are simultaneously sensitive to all parts of the field of view. This multi-element sensitivity is useful for taking several views of an organ after a single radiopharmaceutical administration and has facilitated the dynamic-function studies of more recent interest.

I have found that the design of these collimators is subject to a simple optimization that leads to significant gains in geometric acceptance and thus overall system sensitivity. This optimization is particularly useful because it combines the interdependent factors of resolution and sensitivity.

ANALYSIS

Parallel-hole, multi-aperture collimators are typically designed to provide a certain limiting resolution R at some source-to-collimator distance b. This limiting geometric resolution is defined (*6*) as the full width at half maximum of the average intensity distribution obtained from a point source (Fig. 1) and is given by

$$R = d(a_e + b + c)/a_e, \qquad (1)$$

in which d is the width of the square holes in a rectangular array, c is the collimator-to-detector distance and a_e is the effective length of the holes. The effective length of the holes is less than the geometric length a because the septal material is penetrated by the gamma rays. For single-hole collimators, Mather has shown (*7*) that a_e is given, approximately, by

$$a_e = a - 2\mu^{-1}, \qquad (2)$$

in which μ is the total linear absorption coefficient of the collimator material. It is assumed that this approximation holds for multi-aperture collimators as well.

Anger has shown (*6*) that the geometric acceptance Ω is given by

$$\Omega = [Kd^2/a_e(d + t)]^2 \qquad (3)$$

and is simply the ratio of the number of image-forming gamma rays transmitted by the collimator to those emitted by the object. The symbol t refers to the thickness of the septa in Fig. 1. The constant K depends on the shape of the holes and their distribution pattern and has the value K = 0.282 for square holes in a rectangular array. The septal thickness is chosen so that the transmission for the minimum penetration path W in Fig. 1 is some small percentage such as 5% (i.e., $\exp(-\mu W) = 0.05$, or $\mu W = 3$). This procedure is probably adequate for moderately low resolution collimators for which Ω is relatively large. Penetration path lengths smaller than W do occur at the very edges of the septa, but these are treated semiquantitatively by the Mather approximation (Eq. 2). For most practical collimator designs were $a \gg 2d + t$,

$$t = 2dW/(a - W). \qquad (4)$$

Using the above attenuation criterion,

$$t = 6d/(a\mu - 3). \qquad (5)$$

By substituting Eq. 5 into Eq. 3,

$$\Omega = K^2 \frac{d^2}{a_e^2} \left[\frac{a\mu - 3}{a\mu + 3}\right]^2 \qquad (6)$$

and the explicit dependence on R can be displayed by substitution of Eq. 1, i.e.,

$$\Omega = K^2 \frac{R^2}{(a_e + b + c)^2} \left[\frac{a\mu - 3}{a\mu + 3}\right]^2. \qquad (7)$$

Received Aug. 31, 1967; revision accepted Nov. 9, 1967.

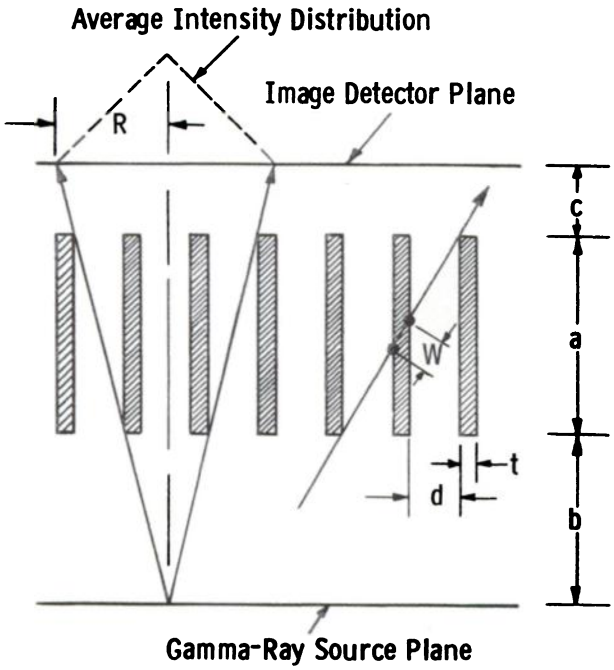

FIG. 1. Sectional view of parallel-hole, multi-aperture collimator showing path of minimum septal attenuation, limiting gamma-ray paths and average intensity distribution on image detector (6).

It is desirable, of course, to maximize the acceptance under the constraints imposed by the desired resolution and the source-detector geometry. A plot of Ω as a function of collimator hole length for several values of R, b = 7.6 cm and c = 0.87 cm (typical for the Anger type of scintillation camera using a 1.3-cm-thick crystal) is shown in Fig. 2. It is clear

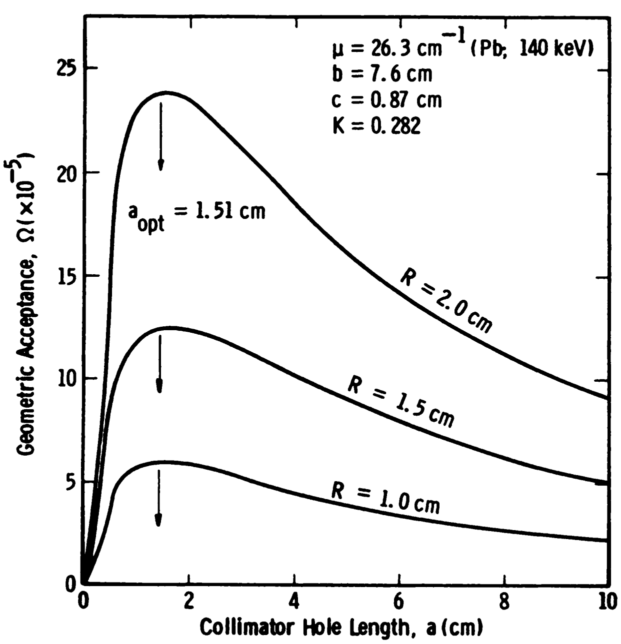

FIG. 2. Geometric acceptance of parallel-hole, multi-aperture collimator versus collimator hole length for constant geometric resolution (Eq. 7).

that an optimum value of Ω exists for each value of R, and it occurs at the same value of a which is designated a_{opt}. This optimum geometric acceptance (Ω_{opt}), found by setting $d\Omega/da = 0$, is readily derived:

$$\Omega_{opt} = \left[\frac{6\mu KR}{(A+6)^2}\right]^2 \qquad (8)$$

where $A^2 = 6 + 6\mu(b + c)$ and

$$a_{opt} = (A + 3)\mu^{-1}. \qquad (9)$$

To satisfy these conditions in the collimator design, the hole width and the septal thickness are chosen so that

$$d_{opt} = R(a_{opt} - 2\mu^{-1})/(a_{opt} - 2\mu^{-1} + b + c) \qquad (10)$$

$$t_{opt} = 6d_{opt}/(\mu a_{opt} - 3). \qquad (11)$$

Table 1 summarizes the optimum parameters for several geometric resolutions and the conditions associated with Fig. 2.

DISCUSSION

There are several features of this optimization, as illustrated in Fig. 2 and Table 1, that should be noted:

1. As illustrated in Fig. 2, Ω_{opt} can be significantly larger than Ω for nonoptimum collimators which provide the same limiting geometric resolution. These differences are important because they relate directly to decreased exposure or integration times (a 50% greater acceptance would result in a 15-min integration being reduced to 10 min). For example, the lead collimator #1 of Table 4 in Ref. 7 has $\Omega = 4.6 \times 10^{-4}$ for 140 kev, R = 1.3 cm and b = 7.6 cm whereas Eq. 8 yields $\Omega_{opt} = 6.7 \times 10^{-4}$, a 47% improvement.

2. The dimensions of optimum collimators tend to have small values and in some cases, for example, with $E_\gamma < 100$ kev and R < 1 cm, special fabrication techniques would be required. For many imaging cases, however, where 140 kev $\lesssim E_\gamma \lesssim$ 364

TABLE 1. OPTIMUM COLLIMATOR PARAMETERS*

$a_{opt} = 1.51$ cm

R(cm)	d_{opt}(cm)	t_{opt}(cm)	$\Omega_{opt}(\times 10^{-4})$
0.30	0.043	0.0070	0.54
0.60	0.087	0.014	2.2
1.0	0.14	0.023	6.0
1.5	0.22	0.036	13
2.0	0.29	0.047	24
2.5	0.36	0.059	38

* For b = 7.6 cm, c = 0.87 cm, $\mu = 26.3$ cm^{-1} (Pb; 140 kev), and K = 0.282.

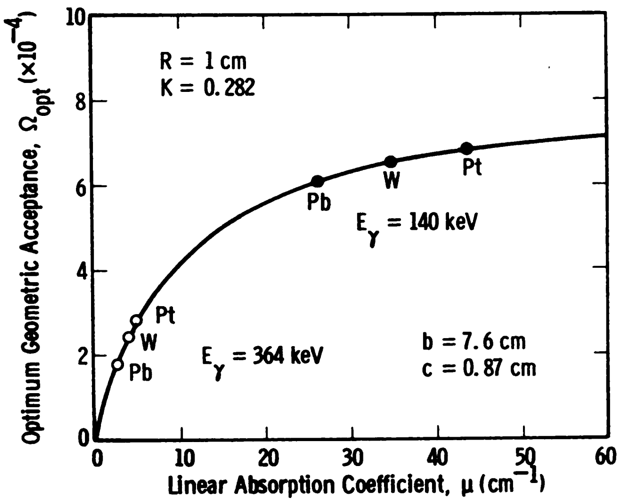

FIG. 3. Optimum geometric acceptance of parallel-hole, multi-aperture collimator versus linear absorption coefficient (Eq. 8).

kev and R > 1 cm, fabrication using thin foil can be readily effected (*3,8*) and has the advantage that denser materials which are difficult to machine in bulk are more easily manipulated (e.g., corrugation).

3. Inspection of Eq. 9 reveals that the optimum length (a_{opt}) is *independent* of the resolution and depends only on b, a characteristic of the organ being imaged, c, a characteristic of the instrument, and μ, a characteristic of the gamma-ray energy (isotope) and the collimator material. The other collimator dimensions, d_{opt} and t_{opt}, are functions of R.

4. The optimum geometric acceptance is relatively *independent* of μ at $E_\gamma = 140$ kev for the collimator materials normally considered for gamma-ray collimators. Equation 8 is plotted in Fig. 3 as a function of μ using the conditions previously considered. Examination reveals that pure tungsten yields an Ω_{opt} only 8% greater than that for pure lead. However, at 364 kev where the coefficients are considerably smaller, tungsten yields a 35% greater acceptance than lead. Hence the material choice is more important at higher energies.

In choosing a collimator for a specific imaging situation, one much realize that R of Eq. 1 is the limiting geometric resolution which corresponds to the spatial frequency $\nu = 1/R$ where the modulation transfer function (MTF) of the collimator is zero. We have shown (*9*) that if a source resolution of ρ is desired, then R should be chosen such that $R \sim 0.4\rho$ to insure that the collimator has an adequate MTF($\sim$60%) at $\nu = 1/\rho$. This requirement is a result of the basic modulation transfer and scintillation limitations (*10*) in gamma-ray imaging and assuming $MTF = (\sin \pi\nu R/\pi\nu R)^2$ as a first approximation for parallel hole, multi-aperture collimators (*9*). Beck and Harper (*11*) have obtained a similar result for focused, multi-aperture collimators used for rectilinear scanning. In addition, the MTF of the gamma-ray image detector should be known so that the entire system can be optimized for over-all resolution and sensitivity.

Finally, it is recognized that gamma-ray cameras are used to image many different organs where a choice of several isotopes for each organ is often available to the clinician. In principle then, an optimum collimator could be designed and fabricated for each organ-isotope combination *and* each resolution desired. Clearly, this is not feasible. One suspects, however, that these various cases can be grouped so that several collimators can serve as nearly optimum for the majority of instances. The performance improvements that could result from this optimization procedure are significant and worthy of implementation by the manufacturers of stationary gamma-ray cameras.

ACKNOWLEDGMENT

I am indebted to John W. Coltman for several stimulating discussions.

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