# Collimation for Imaging the Myocardium. II

L. Stephen Graham, Norman D. Poe, and Gerald D. Robinson, Jr.

*University of California at Los Angeles and Laboratory of Nuclear Medicine and Radiation Biology, Los Angeles, California*

***Line-source response functions and modulation transfer functions (MTF) were used to compare the spatial resolutions obtained with an Anger camera system and four different nuclides used as myocardial-imaging agents: $^{99m}$Tc, $^{123}$I, $^{201}$Tl, and $^{43}$K. The measurements were made with a low-energy converging collimator (LEC), a medium-energy converging collimator (MEC), and a pinhole collimator. The MTF values for $^{99m}$Tc were very similar for all three collimator types, although the LEC collimator gave slightly higher values at high spatial frequencies and had 40% greater sensitivity. Iodine-123 was satisfactorily imaged only with the MEC and pinhole collimators, which in turn yielded MTF values comparable to those measured for $^{99m}$Tc. Thallium-201 produced MTF curves that were similar for the MEC and pinhole collimators; the curve for the LEC collimator was slightly poorer. All three MTF curves for $^{201}$Tl were inferior to those of $^{99m}$Tc. For imaging with $^{43}$K, only the pinhole collimator provided marginally acceptable spatial resolution.***

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Within the last few years, a number of new cyclotron-produced radiopharmaceuticals suitable for imaging regional myocardial blood flow have become available. In general, these agents are simple ions classifiable as potassium analogs and fall into Group I of the periodic table. We have previously reported the physical imaging characteristics of several of these radionuclides, namely, $^{43}$K, $^{81}$Rb, $^{129}$Cs, and $^{13}$N-ammonia (*1*). Subsequent studies have shown that other cyclotron- and reactor-produced radionuclides, which are not Group I elements, are potentially useful for myocardial imaging. These are $^{123}$I [incorporated into fatty acids or other organic molecules (*2*)] and ionic $^{201}$Tl [for regional perfusion imaging (*2–4*)], and $^{99m}$Tc chelates [for infarct localization (*5,6*)].

This report extends the earlier work (*1*) and briefly describes the fundamental advantages and disadvantages of four radionuclides ($^{43}$K, $^{99m}$Tc, $^{123}$I, $^{201}$Tl) from the standpoint of spatial resolution and sensitivity for an Anger camera system. Potassium-43 was included to facilitate comparison of the results of this study with those of the previous one. Xenon-133 was added only because of its similarity to $^{201}$Tl in the energies of their principal photon emissions.

## MATERIALS AND METHODS

Potassium-43 and iodine-123 were produced on site using the following reactions: $^{40}$Ar($\alpha$,p)$^{43}$K and $^{122}$Te(d,n)$^{123}$I. At the end of bombardment, the percentages of contaminants were: for $^{43}$K, 17% $^{42}$K; for $^{123}$I, 0.2% $^{124}$I and 2.6% $^{130}$I. Other impurities from the $^{122}$Te(d,2n)$^{123}$I reaction were not measured; the levels of contaminants present when the experiments were performed were virtually identical to the levels present in commercially available $^{123}$I at calibration. For one set of experiments, $^{123}$I was produced by a $^{127}$I(p,5n)$^{123}$Xe $\rightarrow$ $^{123}$I reaction at the Crocker Nuclear Laboratory, University of Cali-

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For reprints contact: L. Stephen Graham, Nuclear Medicine Div., AR-144B, Center for the Health Sciences, University of California, Los Angeles, CA 90024.

fornia at Davis. Except for $^{125}$I, no other radioactive impurities were identified.

Thallium-201 was generously supplied by the New England Nuclear Corporation (North Billerica, Mass.) and was produced by the reaction $^{203}$Tl(p,3n)$^{201}$Pb → $^{201}$Tl. At the time of calibration, less than 1.5% $^{202}$Tl and less than 0.2% $^{203}$Pb was present, according to data supplied by the manufacturer (*7*). Technetium-99m was obtained from a New England Nuclear $^{99}$Mo–$^{99m}$Tc generator (NRP-196); the levels of contaminants were assumed to be negligible. Xenon-133 gas was obtained from Oak Ridge National Laboratory; we assumed that no radioactive impurities were present.

The line-source response functions (LSRFs) were measured with a Pho/Gamma HP scintillation camera (Searle Radiographics, Des Plaines, Ill.) interfaced to a Hewlett–Packard 5407A scintigraphic data analyzer. All pulse-height windows were set at 20% and were centered on the primary photopeak. The line source had an inside diameter of 1.4 mm and an outside diameter of 1.9 mm. The source was fitted into a 2-mm-square groove in a Plexiglas block, 2.54 cm thick by 40 cm wide by 90 cm long, and a similar block was placed on top of the source. Thus, both absorptive and scattering material was provided. Such a line source is appropriate for myocardial-imaging studies since a large portion of the radioactivity present in the body is outside the field of view. For the low-energy and medium-energy converging collimators, the line source was located 5.2 cm from the collimator face, giving magnification factors of 1.23 and 1.33, respectively. With the line source 10.1 cm from the pinhole collimator face, the magnification factor was 1.64.

After each LSRF measurement, the source was removed from the phantom and a background count was obtained to verify the absence of external contamination. For each nuclide a flood field was used to correct the fields for nonuniformity by a keyboard-initiated program on the scintigraphic data analyzer. Values for the modulation transfer function were calculated directly from the LSRF data using a program modified from Cradduck (*8*).

Due to the small size of the normal myocardium, the three collimators used in the study to provide magnification were: the low-energy Div/Con collimator in the converging mode (Part No. 822017), the medium-energy diverging collimator, inverted for the converging mode (Part No. 821516), and the pinhole collimator with 9-mm aperture (Part No. 820728). In order to invert the medium-energy diverging collimator, an annular ring of lead, 1.2 cm wide and 0.8 cm deep, was removed from the outside edge of the converging side.

## RESULTS

Figure 1 shows MTF values for $^{99m}$Tc, $^{123}$I, and $^{201}$Tl with the low-energy converging collimator (LEC). At all spatial frequencies, $^{99m}$Tc gave the highest MTF values. At most spatial frequencies, $^{201}$Tl had MTF values that were 0.1–0.2 less than the comparable values for $^{99m}$Tc. The MTF curve for $^{123}$I plus contaminants was very poor, having values below 0.2 for spatial frequencies above 0.3 cycles/cm.

To evaluate the full potential of $^{123}$I as a label for imaging the myocardium, MTF values were also obtained for a sample of "pure" $^{123}$I, the only contaminant identified being $^{125}$I, using the LEC collimator (Fig. 2). The MTF values for $^{99m}$Tc were superior to those of "pure" $^{123}$I at all spatial frequencies. At low spatial frequencies (up to 0.3 cycles/cm), the MTF values for "pure" $^{123}$I were less than those of $^{201}$Tl; at higher spatial frequencies, the iodine values were slightly higher (Figs. 1 and 2). Similar measurements were also made for the pinhole collimator; the MTF curves for $^{99m}$Tc and "pure" $^{123}$I were exactly superimposable.

When the medium-energy converging (MEC) collimator was used, the MTF values for $^{99m}$Tc were virtually the same as those obtained with the LEC collimator (Fig. 3). Although the MTF values for $^{201}$Tl were slightly higher with the MEC than with the LEC collimator, they remained essentially the same relative to $^{99m}$Tc. In contrast, marked improvement was seen in the MTF values for $^{123}$I plus contaminants, whose MTF values lay between those of $^{201}$Tl and $^{99m}$Tc. The MTF values for $^{43}$K were below 0.3 for spatial frequencies above 0.2 cycles/cm.

When the pinhole collimator was used, the MTF curves for $^{99m}$Tc and $^{123}$I plus contaminants were virtually identical (Fig. 4). The results for $^{201}$Tl were very similar to those obtained with the LEC and MEC collimators, namely, the MTF values were 0.1–0.2 less than those of $^{99m}$Tc for most spatial frequencies. Although the pinhole collimator substantially improved the MTF curve for $^{43}$K, its MTF was still markedly inferior to those obtained with the other radionuclides. As reported previously (*1*), additional shielding is required to produce satisfactory spatial resolution when $^{43}$K is used for imaging the myocardium.

Line-source response functions were also measured and modulation transfer functions calculated for $^{133}$Xe. The only substantial difference between the values calculated for $^{201}$Tl and $^{133}$Xe was noted for the LEC collimator. In that case, the MTF values for $^{201}$Tl were 0.02–0.05 less for spatial frequencies of 0.05–0.45 cycles/cm.

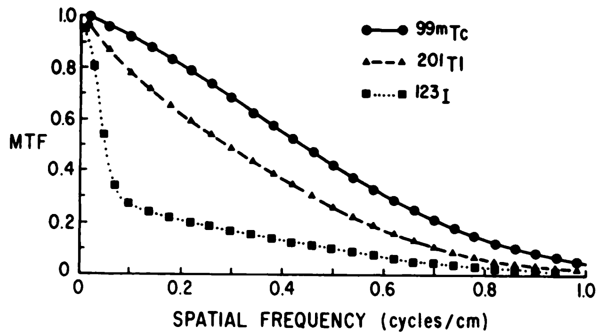


**FIG. 1.** Modulation transfer functions for $^{99m}$Tc, $^{123}$I, and $^{201}$Tl using Anger camera with low-energy converging collimator. (See text for specification of radionuclidic purity of $^{123}$I solution.)

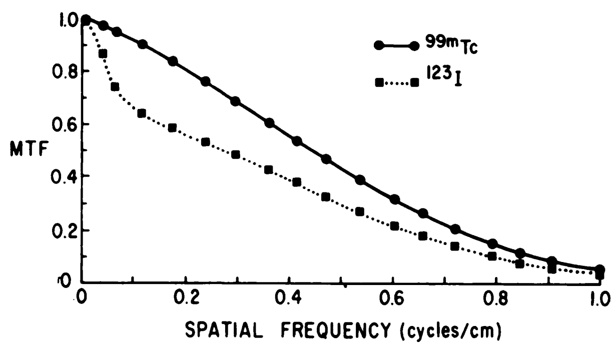


**FIG. 2.** Modulation transfer functions for $^{99m}$Tc and pure $^{123}$I using Anger camera with low-energy converging collimator.

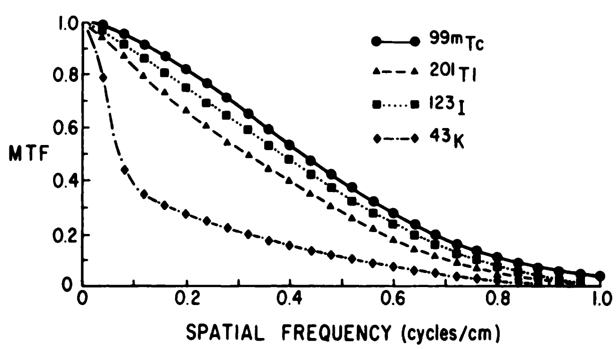


**FIG. 3.** Modulation transfer functions for $^{99m}$Tc, $^{123}$I, $^{201}$Tl, and $^{43}$K using Anger camera with medium-energy converging collimator.

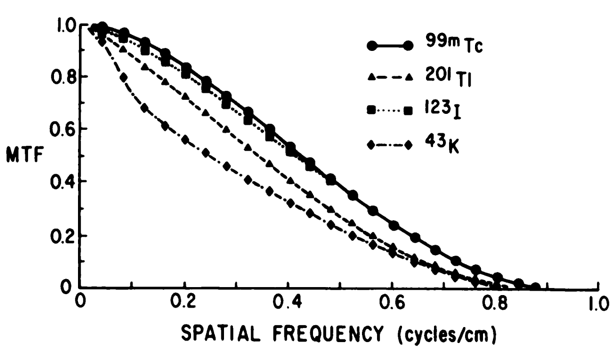


**FIG. 4.** Modulation transfer functions for $^{99m}$Tc, $^{123}$I, $^{201}$Tl, and $^{43}$K using Anger camera with pinhole collimator.

Since some contamination was present in the $^{201}$Tl solution, LSRFs were obtained and MTFs calculated daily for up to 6 days after the initial measurement. By the sixth day, some loss of spatial resolution was reflected in the MTF curve for the LEC collimator. However, the difference was never more than 0.05 at any spatial frequency. No difference was noted in the MTF curves for the pinhole or the MEC collimator as a function of time.

To study the detector efficiencies for each nuclide, the appropriate photon yields were multiplied by the photopeak efficiency of a 0.5-in.-thick NaI(Tl) crystal for the corresponding photon energies. The results are given in relative form in Table 1. Based on these calculations, $^{201}$Tl had a detector efficiency 32% better than that of $^{99m}$Tc, whereas $^{123}$I detection was about 13% poorer than that of $^{99m}$Tc. Potassium-43 was markedly lower than any of the other nuclides.

The relative collimator sensitivities were evaluated using the integrals of the appropriate line-source response functions (Table 2). The relative sensitivities for the MEC and pinhole collimators were essentially the same for all of the nuclides except $^{43}$K. For $^{99m}$Tc and $^{201}$Tl, the LEC collimator had a relative sensitivity that was 40–50% greater than the pinhole collimator. The higher sensitivity with $^{123}$I reflects excessive penetration of the collimator septa by high-energy photons and is associated with poor spatial resolution.

## DISCUSSION

Although a wide variety of nuclides are currently used in nuclear medicine studies, most collimators have been designed for either $^{99m}$Tc or $^{131}$I. Progress in collimation for $^{99m}$Tc has led to the development of foil collimators with thin septa; these provide high spatial resolution while maintaining moderate sensitivity. Septal thicknesses of 0.25 mm are adequate for the monoenergetic radiations from $^{99m}$Tc and other nuclides that emit lower-energy photons, but even small abundances of high-energy photons may cause a serious loss of spatial resolution for collimators developed for $^{99m}$Tc. On the other hand, any increase in septal thickness will reduce collimator efficiency. Accordingly, in selecting a collimator for a specific radionuclide, it is imperative that the nuclide be carefully evaluated with respect to the spatial resolution and sensitivity that a given collimator will provide.

If an agent used for myocardial imaging can be labeled with $^{99m}$Tc, the spatial resolution obtained will be superior to that of any of the other nuclides studied in these experiments. This fact, coupled with high sensitivity (Table 1), makes $^{99m}$Tc the nuclide

**TABLE 1. RELATIVE CRYSTAL EFFICIENCY**

| Nuclide | Principal photon energies* (keV) | Abundance (%) | Half-life (hr) | Relative photopeak efficiency for 0.5-in. NaI(Tl)† |
|---|---|---|---|---|
| $^{99m}$Tc | 140 | 87.9 | 6.03 | 1.0 |
| $^{123}$I | 159 | 83.6 | 13.0 | 0.87 |
| | 347–784 | 2.2 | | |
| $^{201}$Tl | 69–83 | 98 | 74 | 1.32 |
| | 135, 167 | 10 | | |
| $^{43}$K | 373–394 | 98.9 | | |
| | 592–617 | 80.9 | 22.4 | 0.25 |

* Data taken from Refs. *2* and *9*.

† The abundance of $^{99m}$Tc photons multiplied by the photopeak efficiency for a 0.5-in.-thick NaI(Tl) crystal was arbitrarily taken as 1.0. The tabulated values were calculated using data in Ref. *10*.

**TABLE 2. RELATIVE SENSITIVITIES***

| Nuclide | Collimator: Low-energy converging | Medium-energy converging | Pinhole |
|---|---|---|---|
| $^{99m}$Tc | 1.4 | 1.0 | 1.0 |
| $^{201}$Tl | 1.5 | 1.0 | 1.0 |
| $^{123}$I (Te target) | 4.1 | 1.1 | 1.0 |
| $^{123}$I (I target) | 2.3 | 1.0 | 1.0 |
| $^{43}$K | | 1.8 | 1.0 |

* Determined separately for each nuclide by assigning a value of 1.0 to the pinhole collimator.

of choice from the standpoint of physical characteristics. The spatial resolution potentially available through the use of $^{99m}$Tc-labeled radiopharmaceuticals and the LEC collimator is slightly better at high spatial frequencies than that obtained using the pinhole collimator. This improvement would probably not be clinically observable and is seen only because of the extreme sensitivity of the modulation transfer function as a measure of spatial resolution. For $^{99m}$Tc, the MTF curves for the LEC and MEC collimators were essentially identical. However, the relative sensitivities of the three collimators for $^{99m}$Tc [namely, 1.4 (LEC), 1.0 (MEC), and 1.0 (pinhole)] indicate that the LEC collimator would be the best choice (Table 2).

When the MTF curves for $^{201}$Tl are compared for different collimators, the results are somewhat different. The MTF values for the LEC collimator were slightly inferior to those obtained with the MEC collimator. That result probably reflects the effect of scatter into the 80-keV window from the higher-energy gamma photons present in $^{201}$Tl (135 keV and 168 keV) and also the presence of contaminants. As Figs. 1 and 3 show, the difference was very small. While the pinhole collimator gave slightly better spatial resolution for the specific experimental conditions used with $^{201}$Tl, its low sensitivity would not make it the collimator of choice. In fact, the slight improvement in spatial resolution may have been due to the greater magnification factor.

Compared to $^{99m}$Tc, the MTF curves for $^{201}$Tl were inferior in every case. That result is to be expected because of the poorer intrinsic resolution of Anger camera systems for photons of lower energy, at least up to the point where multiple scattering in the scintillator becomes an appreciable factor (*11*). Although the difference in spatial resolution should be observable in a static imaging situation, the motion of the myocardium may partially offset the inherent advantage of $^{99m}$Tc.

Although the calculated relative crystal efficiency of $^{201}$Tl (1.32) is considerably higher than that of $^{99m}$Tc (1.00), the full advantage of this high efficiency cannot be realized in practice. Because the range of x-ray energies extends from 60 to 83 keV, a 20% window cannot cover the photopeak. A wider window would improve the efficiency but would also increase the amount of scattered radiation accepted by the pulse-height analyzer. Inclusion of increased amounts of scattered radiation would further reduce the spatial resolution.

When substantial amounts of $^{124}$I and $^{130}$I are present in $^{123}$I solutions, the LEC collimator cannot be used for imaging the myocardium (Fig. 1). Although the high energies (511–1,691 keV) and high abundance (133.1%) of gamma photons from $^{124}$I lead to a loss in spatial resolution, the higher abundance (324.8%) of the high-energy photons (418–1,157 keV) from $^{130}$I produces even greater degradation for the same amounts of activity. The excessive penetration of the septa by the high-energy photons from both of these nuclides, as well as the high-energy component from $^{123}$I itself, are reflected in the "high sensitivity" of the LEC collimator (Table 2). Of course, this high sensitivity is of no clinical value because of the associated poor spatial resolution. The MTF curve presented in Fig. 1 was calculated from a LSRF measurement made approximately 6 hr after the end of bombardment; it represents a purity (for the production method specified) that can be realized only if the cyclotron is on site.

It is extremely important to note that even when "pure" $^{123}$I is used, the LEC collimator is not satisfactory (Fig. 2). Although the results may seem surprising in light of the low photon abundance at high energies (Table 1), one must realize that the thin septa do not effectively attenuate 347–784-keV photons. Thus, photons may approach the crystal from many different angles, as shown by the relative increase in sensitivity of the LEC collimator compared to that of $^{99m}$Tc (2.3 vs. 1.4).

The abundance of high-energy photons from $^{43}$K makes collimator design for that nuclide a difficult problem. As was shown previously, the spatial resolution was very poor when the higher-energy photons (592–617 keV) were used for image formation (*1*). Even when a 20% window was set for the lower range of photon energies (373–394 keV), the MEC collimator was still unsatisfactory (Fig. 3). Although the pinhole collimator was superior to the MEC collimator in terms of spatial resolution, the MTF values for $^{43}$K were still 0.1–0.3 below those obtained for $^{99m}$Tc (Fig. 4). Additional shielding does improve the images obtainable with $^{43}$K and the pinhole collimator (*1,12*), but the low efficiency (Table 2) probably does not justify the extra expense.

## CONCLUSION

For studies where magnification factors of 1.2–1.3 are desirable, the low-energy converging collimator provides the best spatial resolution and sensitivity for both $^{99m}$Tc and $^{201}$Tl. Nevertheless, all else being equal, $^{201}$Tl cannot provide as high a spatial resolution as may be obtained with $^{99m}$Tc because of the poorer intrinsic resolution of thallium in Anger camera systems. For $^{123}$I produced from a $^{122}$Te target, the medium-energy converging collimator gives MTF values that are only slightly inferior to those of the pinhole collimator; both collimators have about the same sensitivity for the experimental conditions used in this work. "Pure" $^{123}$I cannot be satisfactorily imaged with collimators optimally designed for $^{99m}$Tc because of the abundance of high-energy photons. For high-resolution images the medium-energy converging or the pinhole collimator should be used. When $^{43}$K is used for imaging the myocardium, only the pinhole collimator provides adequate spatial resolution. Improved images can be obtained with $^{43}$K by using additional shielding.

When viewing the images produced by Anger camera systems with magnifying collimators, one should remember that the amount of magnification is a sensitive function of depth. This variable magnification causes distortions that must be considered when the resultant image is interpreted (*13,14*).

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