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A Line-Source Phantom for Testing the Performance of Scintillation Cameras

Eberhard Jahns[1] and Gerald J. Hine[2]

Heidelberg, Germany and Vienna, Austria

INTRODUCTION

The performance of scintillation cameras has been described recently by several investigators (1, 2, 3). Their determinations of the camera characteristics are based on elaborate and time consuming measurements which are not practical for routine testing of many instruments. Furthermore, the performance of a camera should be checked at each laboratory not only during its installation but at frequent intervals in order to secure optimum operating conditions at all times.

With a simple qualitative method, to be described, the principal characteristics of a scintillation camera can be verified in a short time. They include the following:

1. Uniformity of camera performance for the whole field of vision.
2. Spatial resolution of scintillation camera for γ rays of various energies.
3. Depth response of scintillation camera for sources in air and in absorbing medium (water).
4. Effect of operating conditions on camera performance.

METHOD

A geometrical phantom[3] (Fig. 1) has been constructed that is easy to reproduce. With a circular saw grooves are cut into a 10 mm thick rectangular Lucite plate 30- by 30-cm wide. Parallel to one side, the grooves are 3-mm deep, while those parallel to the perpendicular side are about 6-mm deep. The grooves form a pattern of 5-cm squares with additional grooves 2.5- and 1.25-cm apart at the center of the phantom.

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[1]Universitäts-Strahlenklinik (Czerny-Krankenhaus), Heidelberg, Fed. Rep. Germany.

[2]Section of Medicine, Department of Research and Isotopes, International Atomic Energy Agency, Vienna, Austria.

[3]Available from Strahlentechnik GmbH, Deutscher Picker-Vertrieb, Hamburg, Fed. Rep. Germany.

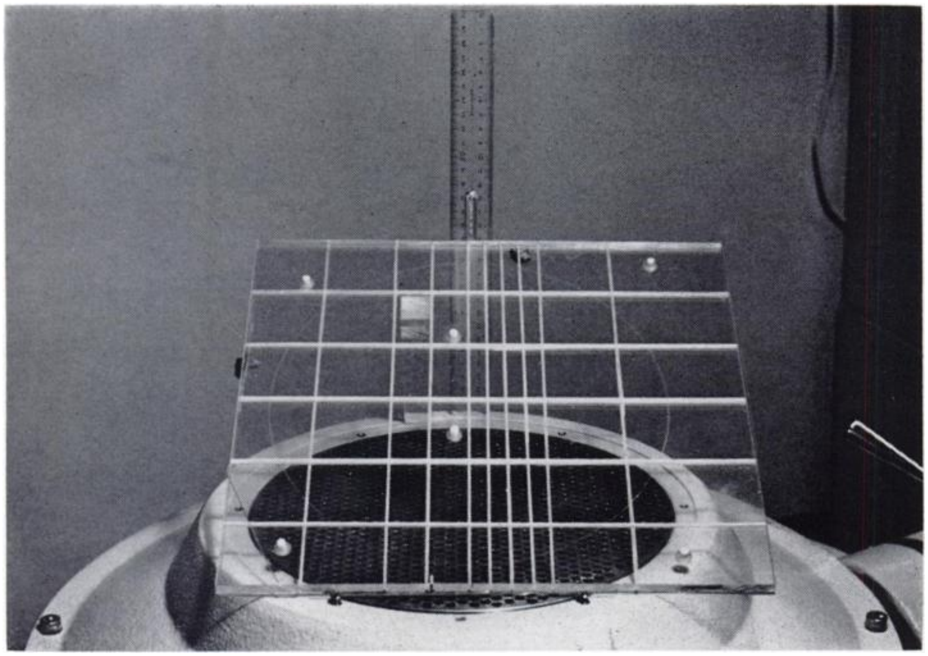

Fig. 1. Line-source phantom in tilted position on top of multichannel collimator of scintillation camera. Thin plastic tubes containing a radioactive solution are placed in grooves of Lucite phantom.

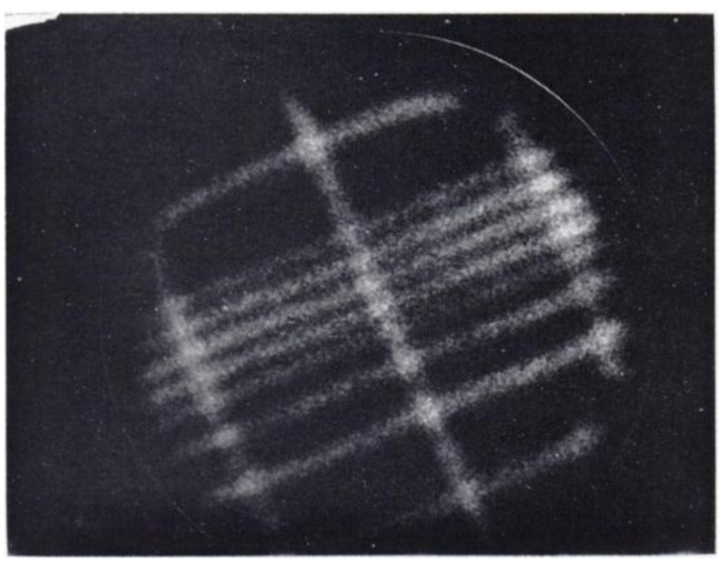

Fig. 2. Scintiphoto of line-source phantom in contact with face of collimator. Using the ^{203}Hg γ rays, 10^5 counts were collected. Uniformity of scintillation camera performance for most parts of the field of vision is satisfactory.

The surface of the Lucite plate containing the grooves is covered by a 3-mm thick sheet of Lucite. A thin plastic tube with less than 2-mm outside diameter is threaded through the grooves by passing on the outside from one groove to another. According to the intended use of the phantom, an appropriate pattern of tubes may be chosen. After injection of a radioactive solution with a syringe, both ends of the plastic tube are closed tightly.

The phantom may be positioned at an angle with respect to the face of the multichannel collimator (Fig. 1), or it may be placed parallel to it at various distances. The effect of an absorbing medium between source and collimator can be determined by putting the phantom into a tank with water.

The line-source phantom, as viewed by the scintillation camera, is depicted as scintiphotos that are Polaroid pictures of the light flashes on the oscilloscope screen. This fast method of data recording allows only a qualitative interpretation of the results, as exemplified by clinical data obtained with the scintillation camera. More quantitative information could be obtained by using a digital readout method (2).

RESULTS

The usefulness of the line-source phantom will be illustrated by the results obtained with a Nuclear-Chicago scintillation camera.[1] A multichannel lead collimator with 1165 holes is employed. Each hole has a diameter of 0.237 inch and a length of 3.0 inches.

Uniformity of response and resolution of camera

The scintillation camera should reproduce a radioisotope distribution without distortions, which means that linear sources should appear as straight lines and the angles between line sources should be preserved. This is by no means a trivial requirement for the performance of a scintillation camera and it must be verified.

The line-source pattern of Figure 2 was obtained using ^{203}Hg as γ-ray emitter. The phantom was placed in contact with the multichannel collimator and the line sources were oriented parallel to the rows of holes in the collimator. The appearance of the line-source pattern in Figure 2 is fairly uniform; the sources appear as straight lines and at right angles to each other, but in some areas they seem to have less radioactivity. Only at the outer edge a barrel-like aberration is noticed. Similar results have been obtained by Anger (4) using different types of test patterns.

Freedom from distortions alone does not guarantee a high quality of reproduction. For delineating adjacent differences in radioisotope content within an organ, the spatial resolution of the camera must be adequate and constant for all sections of its sensitive area. Using the line-source phantom, the resolution of the scintillation camera can be checked simultaneously with the uniformity of its performance.

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[1]Nuclear-Chicago Corp., "Pho/Gamma" Scintillation Camera.

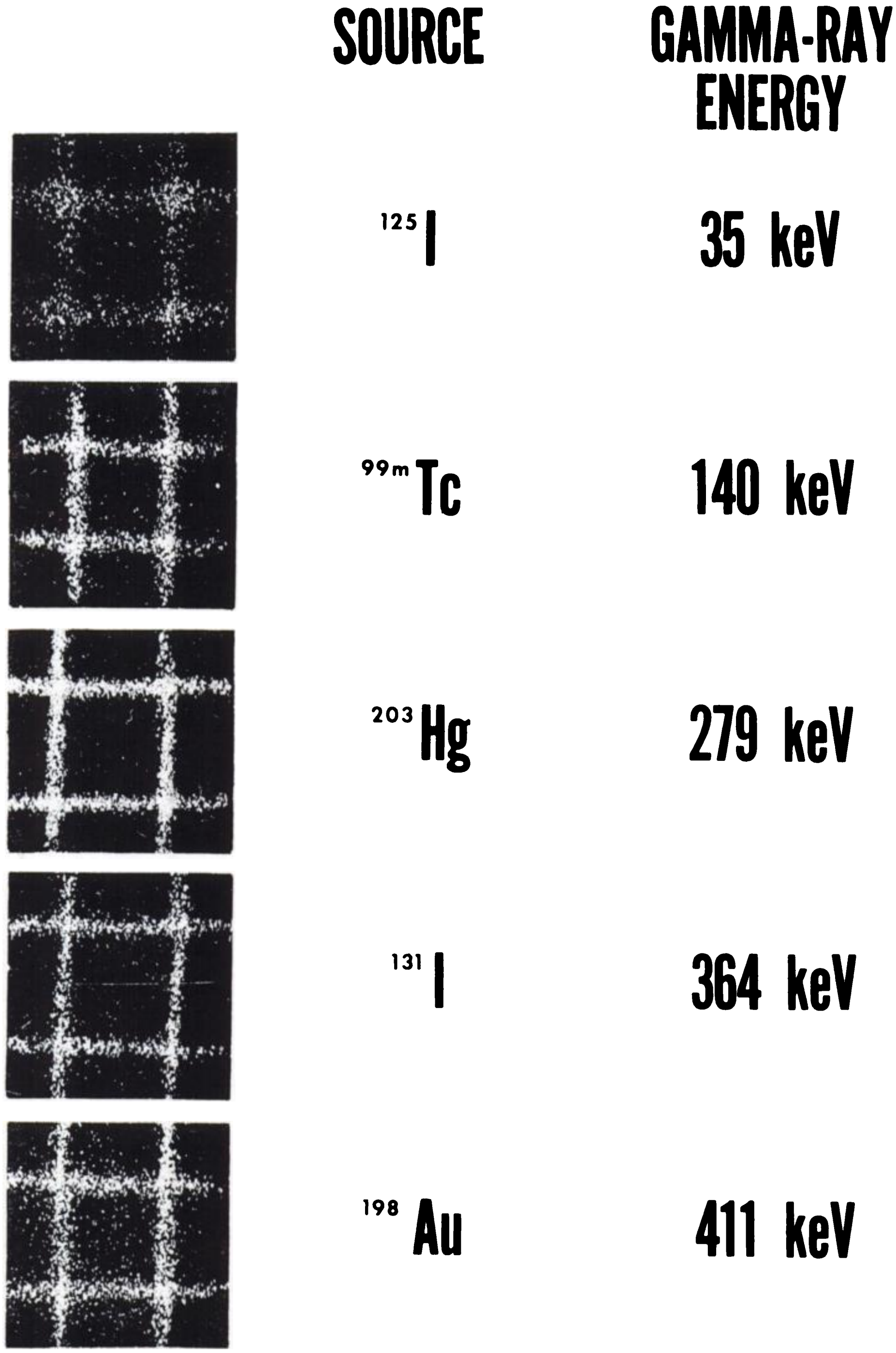

Fig. 3. Dependence of spatial resolution of scintillation camera on γ-ray energy. Line-source phantom with only four sources, 10 cm apart, in contact with 1165-hole collimator, yielded best results with ^{203}Hg sources.

In Figure 2 the line sources which were 2.5 cm or more apart appear clearly distinct from each other. Those separated by only 1.25 cm can still be recognized as separate sources, but not equally well at all sites. When the line sources were not aligned with the rows of holes in the collimator, an interfering line structure was observed which was caused by the multichannel collimator. This phenomenon indicates that the intrinsic resolution of the scintillation camera (without collimator) is superior to the one observed with the present collimator (1, 2).

The uniformity of response and spatial resolution of the scintillation camera may change with time, for example, because of temperature variations. Therefore, it should be checked at regular intervals. In our experience, a weekly test is sufficient for the instrument at our disposal.

Dependence of camera resolution on γ-ray energy

The spatial resolution of the scintillation camera is affected by the energy of the γ rays for two reasons. First, the collimator material is not impervious, especially to higher energy γ rays. For the present collimator, the lead septa between the holes are 0.075 inch thick. Second, the amount of light produced per γ ray absorbed in the sodium iodide crystal is proportional to its energy. With decreasing γ-ray energy, the uncertainty for localizing the origin of the light in the scintillator increases because of the reduced number of light quanta available for each multiplier phototube. Therefore, although for different reasons, a poorer performance of the scintillation camera must be expected as the energy of the γ rays deviates in either direction from an optimum value.

Figure 3 shows the scintiphotos of four line sources with the phantom forming a 10- by 10-cm square taken with five γ-ray emitters. The phantom was placed again in contact with the multichannel collimator. The sequence of pictures reveals the two effects described above; with increasing γ-ray energy, up to about 300 keV, the sources appear better defined. Because of septa penetration, a uniform background appears which is enhanced with increasing γ-ray energy and furthermore the sources are again broadened at the energy of the ^{198}Au γ rays. The best overall picture is obtained with the ^{203}Hg γ rays. This result is in agreement with recent observations (1, 2, 4) which find the spatial resolution of the scintillation camera with the ^{203}Hg γ rays superior to that with the higher-energy γ rays of ^{131}I.

Depth response of camera

While the camera performance described above was determined with the line-source phantom in contact with the collimator, in clinical practice the organs containing the radioisotopes are always at some distance from the collimator. Therefore, changes of camera performance must be in investigated for various source-to-collimator distances. For this purpose, the line-source phantom may be positioned either at an angle with respect to the collimator as shown in Figure 1, or it may be placed parallel to the collimator face at various distances from it.

The results obtained with the phantom in a tilted position are shown in Figure 4. The sources which were only 1.25 cm apart appear distinct from each other to a depth of about 6 cm, while at greater depths they can no longer be separated. This observed change in camera resolution with depth is in agreement with calculations by Anger (1) and measurements by Myers *et al* (2).

With the line-source phantom parallel to the face of the collimator, the effect of varying the source-to-collimator distance on the camera performance can be investigated in more detail. In Figure 5, a series of scintiphotos is shown that has been obtained with a preset number of dots (10^4) per exposure. With the phantom in air, the exposure time (t) is practically constant; the sensitivity of the scintillation camera does not depend on the distance between the line-source phantom and the collimator ($d = 1$ to 20 cm). With water as an absorbing medium between source and collimator, analogous to the conditions within the body, the exposure time increases exponentially with increasing absorber thickness ($t = 27$ to 188 sec).

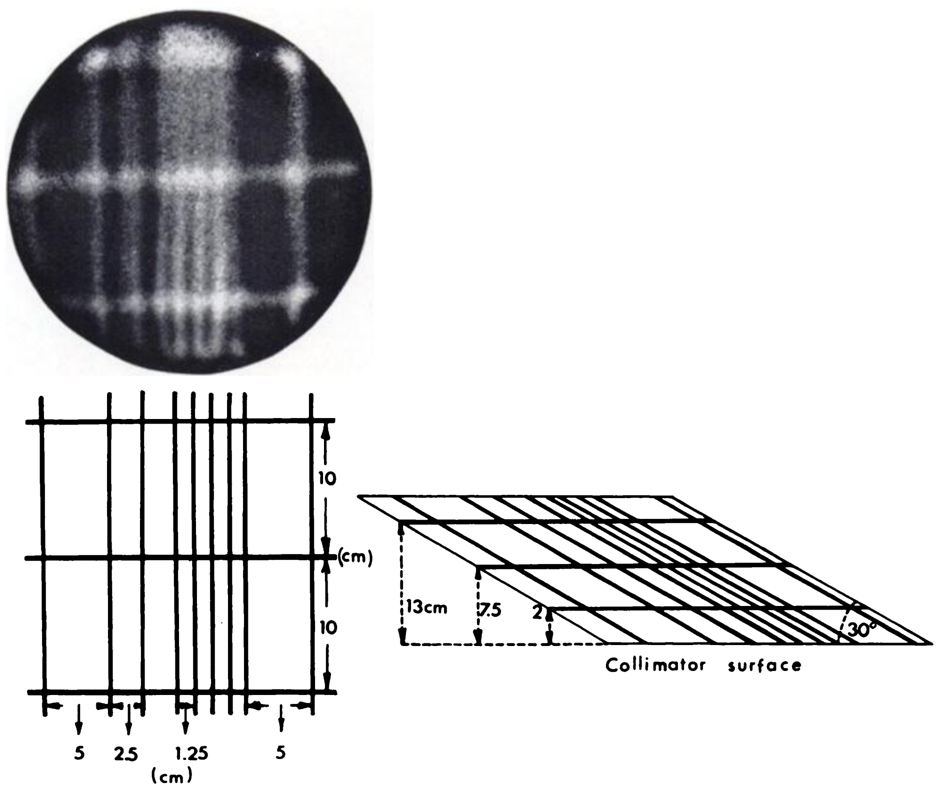

Fig. 4. Test of depth response of scintillation camera with line-source phantom in tilted position as shown in Fig. 1. Results obtained with ^{203}Hg γ rays verify expected camera performance.

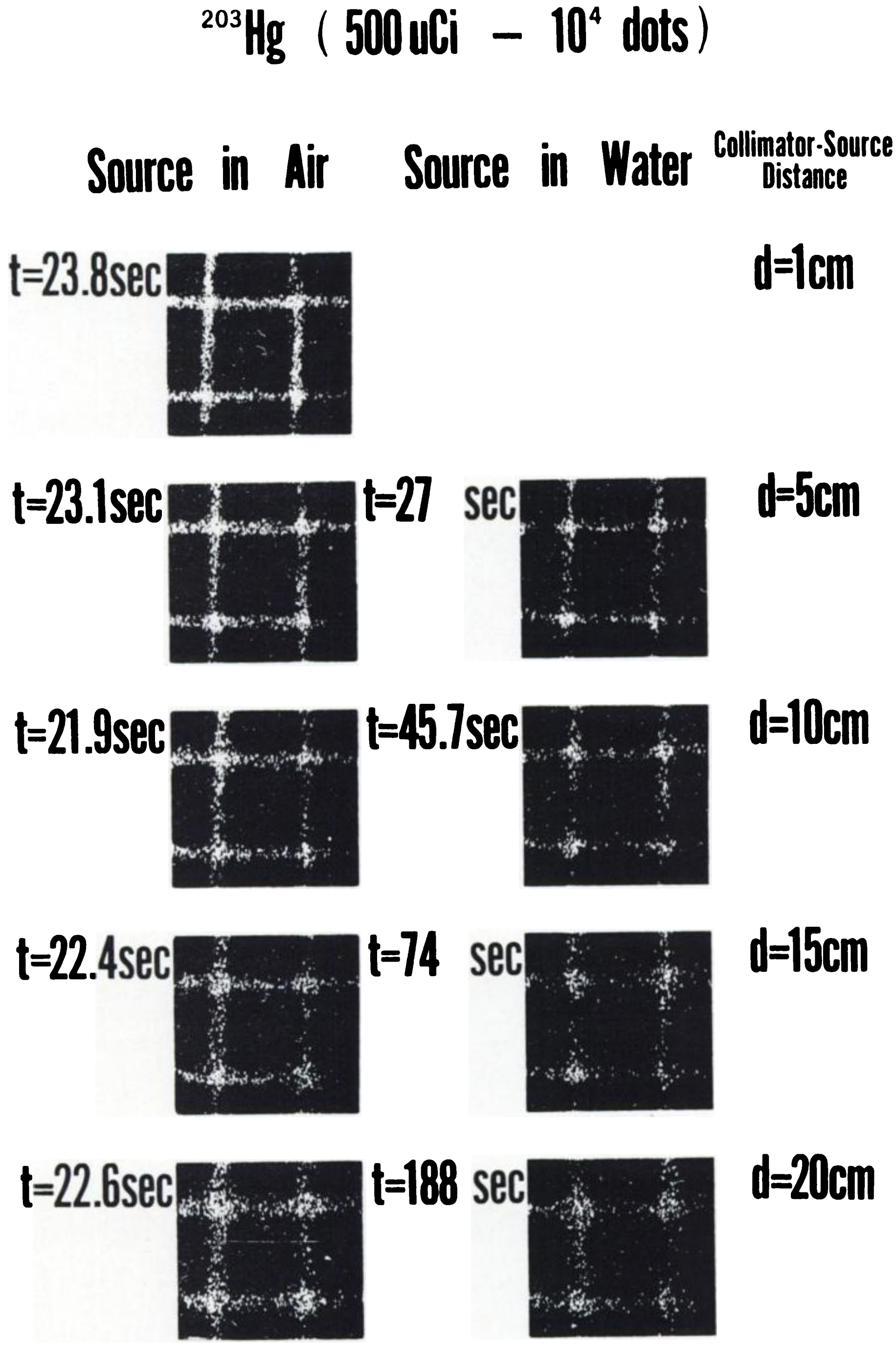

Fig. 5. Scintiphotos with line-source phantom parallel to collimator face at several distances, *d*, in air and in water. For the ^{203}Hg sources in air, the sensitivity of the scintillation camera is constant for all source-to-collimator distances while the spatial resolution decreases with increasing distance. With the sources in water the exposure time, *t*, for a constant number of dots increases rapidly because of γ-ray absorption.

Finally, for the sources in air, Figure 5 shows again the decrease in spatial resolution with increasing source-to-collimator distance noticed in Figure 4. As the line sources are moved away from the collimator, they appear broadened on the scintiphotos. In water, the width of the sources will be further increased slightly because of small angle scattering within the absorbing medium. The change in spatial resolution with distance, however, is about the same as that observed in air.

Operating conditions for camera

For the tests of the camera performance described above, the camera was operated in the same way as for clinical investigations. In particular, the width of the spectrometer window which determines the fraction of scintillation pulses used for the data analysis was set to encompass only the center section of the photopeak; the full width of the window was chosen as 10% of the peak energy. Though this yields the best camera performance, it reduces its sensitivity.

In order to eliminate unnecessary statistical fluctuations, a higher number of counts has been used than will be practical in many medical applications. Therefore, clinical scintiphotos with a smaller number of dots may appear less well-defined, although the camera performance is not affected by the total number of counts that are accumulated.

The line-source phantom appears well suited for investigating the above and other problems. It should be used for establishing the best operating conditions for a particular combination of multichannel collimator, γ-ray emitter, width of spectrometer window and clinical problem requiring a certain resolution, total number of counts, etc.

SUMMARY

By means of a simple line-source phantom, the principal characteristics of a scintillation camera can be verified in a short time. The uniformity of camera performance, its spatial resolution and its depth response may be checked for either a single or several γ-ray energies. Scintiphotos obtained with a Nuclear-Chicago "Pho/Gamma" scintillation camera are presented. The phantom may be used for routine checking of any camera system as well as for determining the best operating conditions for a particular clinical problem.

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