A Phantom for Clinical Evaluation of Total System Resolution

Gary W. Enos*

Holy Family Hospital, Spokane, Washington

A phantom designed to simulate clinical parameters was constructed and evaluated to provide a more practical evaluation of Anger system resolution. The phantom permits subjective evaluation of spatial resolution at different depths, in the presence of scattering medium and varied ~~input~~ contrast.

The intrinsic spatial resolution specifications provided by Anger camera system manufacturers usually reflect the most favorable experimental circumstances (*1*). Intrinsically, current models are capable of resolving high contrast objects separated by 2.5 to 3.0 mm in the energy region of 140 keV (*1*). A more realistic assessment of the resolving power is obtained with the presence of sufficient scattering material to simulate clinical conditions (*1*).

There are a multitude of methods available for evaluating a number of parameters—including uniformity, linearity (X and Y), spatial resolution, count efficiency, and image distortion—in the Anger camera. In the majority of cases, these evaluations are made under other than clinical conditions. Commonly used phantoms utilize lead bars or holes drilled in lead (standard bar phantoms, Smith orthogonal hole phantoms [ADC Medical, Farmingdale, N.Y.] to provide high-input contrast to the detector assemblies. These methods usually use ¼-in. thick lead representing 28 half-value layers for 140-keV photons, or ¼-in. powered tungsten representing 32 half-value layers for 140-keV photons. Although the fidelity increases as bar width and interspacing decreases, lack of a scattering medium and presentation to the detector of a high-contrast imput tend to represent other than the usual clinical problem—namely, the detection of a focal void within an activity distribution (*2*).

Line spread function measurements, made with line sources placed in air or scattering media, are difficult to perform and do not reflect the clinical problem. The most popular technique, involving determination of the width of the curve at half the maximum value (FWHM) (*3*), does not reflect the effect of scattering medium, as deterioration of image quality because of this clinical problem is rarely seen above the 50% level in this measurement (*1*).

Results of a modulation transfer function measurement yield data relating primarily to the spatial resolution performance on a sinusoidal distribution of radioactivity. The method measures how well an imaging device can represent object contrast of a given spatial frequency into image contrast. Again, the modulation transfer function is difficult to perform, and the data do not reflect the imaging device's response to detection of voids under clinical parameters (*2*).

Contrast efficiency can be evaluated by using the Rollo phantom (Nuclear Associates, Inc., Carl Place, NY). This phantom, constructed from Lucite, contains an inner core that measures 20.3 × 20.3 × 7.62 cm, and contains 16 interconnecting cells. The cells are filled with a radioactive solution and four different sphere diameters at four different contrast levels are imaged within 7.5 cm of Lucite scattering medium (*2*). Images of the phantom obtained on different collimators for equal data accumulation times will reflect trade-offs between contrast efficiency and sensitivity offered by each collimator type (*2*). The principal disadvantage of the Rollo phantom is that it only offers evaluation of the imaging device's central portion, unless multiple images across the detector assembly are obtained. Evaluations of linearity and image distortion over the entire detector assembly are not easily appreciated.

With the preceding information in mind, I developed a phantom that evaluates the total system Anger camera performance under clinical parameters.

Materials and Methods

The phantom, filled with water to simulate clinical scattering, is constructed of Lucite 0.635-cm thick and measures 32.4 × 32.4 × 7.62 cm. Lead disks, 0.79-mm thick, are placed on five 1.0-cm interval steps within the phantom at depths of 0.0–4.0 cm (3.0–7.0 cm when the phantom is inverted). Each step displays the disks in groups of five, with diameters increasing by a 3.2-mm interval—the smallest set has a 3.2-mm diameter and the largest set a 19.0-mm diameter. Each group of five equally-sized disks is patterned to evaluate spatial resolution. Each disk represents three half-value layers of attenuation for 140-keV photons, a condition providing an initial object contrast of 0.875. With addition of water to the phantom and use of a transmission source, the effects of scattering medium and poor geometry reduce this value to 0.5. With addition of varying quantities of technetium to water within the phantom, object contrasts can be reduced to unlimited values (Fig. 1).

We only evaluated the phantom on those Anger camera systems specifying an intrinsic resolution of 3.2 mm or

*Present address: Picker Corp., 12 Clintonville Rd., Northford, CT 06472.

For reprints contact: Gary W. Enos, Picker Corp., 12 Clintonville Rd., Northford, CT 06472.

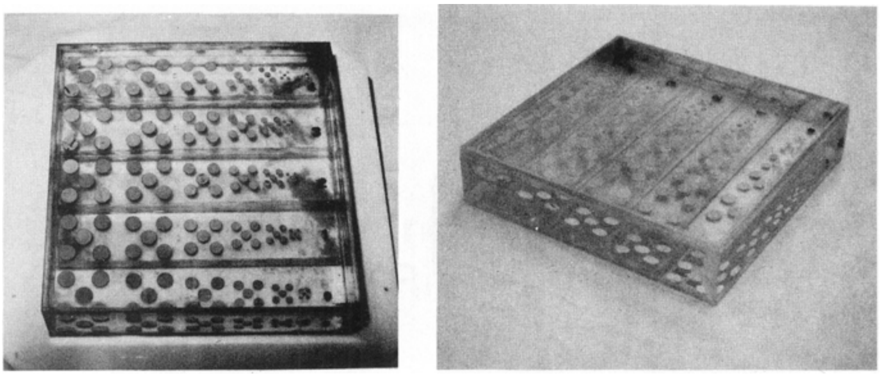

Fig. 1. Demonstration of the phantom disk patterns and the stepped depth configuration described from different perspectives.

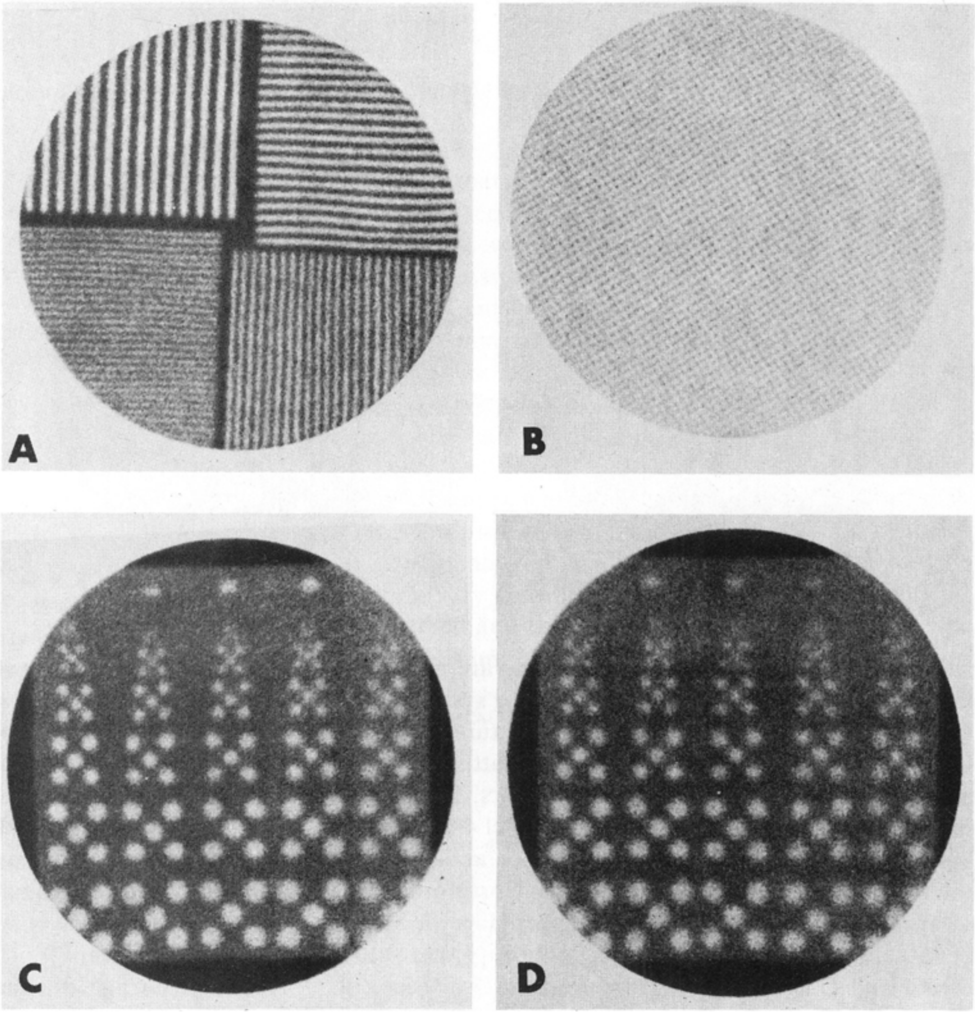

Fig. 2. (A) Standard bar phantom (1/4 in., 3/16 in., 1/8 in., 5/32 in.) obtained on the surface of the collimator of the Picker system. (B) Smith orthogonal hole phantom image obtained intrinsically (1/8 in. holes, 1/4 in. centers). (C) Phantom described imaged extrinsically representing 0.0-cm through 4.0-cm depth and (D) 3.0-cm through 7.0-cm depth.

better. Comparable 3,000 counts per square cm ID images were obtained at 15% windows for 140-keV photons on camera systems equipped with ultrafine high-resolution, low-energy collimators. The counting rates at which phantom images were obtained did not exceed 7,000–8,000 counts per sec (most clinical images are performed at a rate of 1,000–10,000 counts per sec) (*1*). We used Kodak NMB film (Eastman Kodak Co., Rochester, NY) to obtain images on integral multiformatting systems. For reference, intrinsic ⅛-in. Smith orthogonal phantom images and extrinsic standard bar phantom images were obtained, in addition to the extrinsic images of the phantom described. When a uniformity correction device was a part of the system, the images obtained were with this unit in operation.

Utilizing the phantom, adjustments were made on dot brightness and *f*-stop settings to assure proper exposure levels. The effects of minor adjustments on maximizing output contrast levels are easily appreciated with this phantom.

Results and Summary

To illustrate the application of the phantom, images

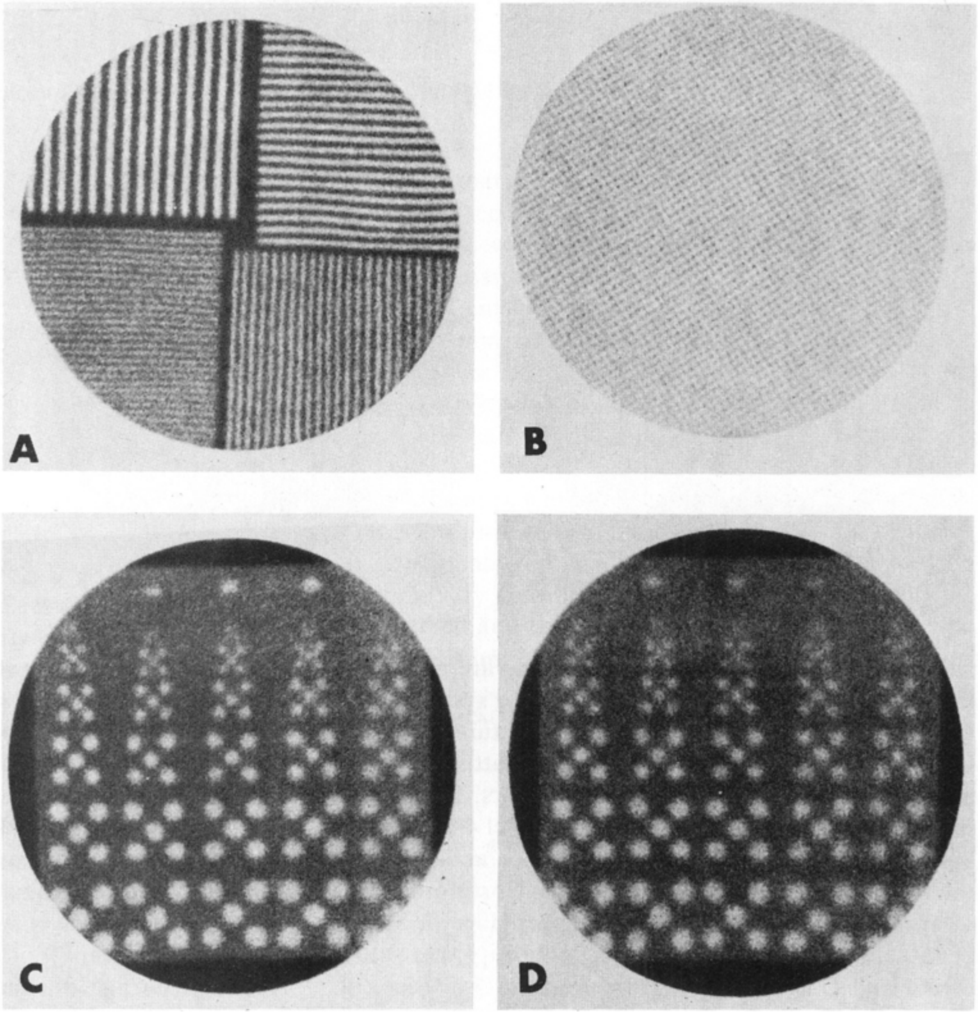

Fig. 3. Standard bar phantom (1/4 in., 3/16 in., 1/8 in., 5/32 in.) obtained on the surface of the collimator of the General Electric system. (B) Smith orthogonal hole phantom image obtained intrinsically (1/8 in. holes, 1/4 in. centers). (C) Phantom described imaged extrinsically representing 0.0-cm through 4.0-cm depth and (D) 3.0-cm through 7.0-cm depth.

from two 1978 model large-field cameras are presented (Figs. 2 and 3). In both cases, the imaging systems were equipped with original equipment manufacturer uniformity correction devices, 8 × 10 in. multiformatting systems, and high-resolution, low-energy collimators. The images in Fig. 2 were obtained on a Picker model 4-15 camera system equipped with Micro-Z processor and a Picker multiformatting system (Picker Corp., Northford, CT). Images in Fig. 3 were obtained on a General Electric Maxi-II camera system equipped with uniformity correction device and G.E. multiformatting system (G.E. Medical Systems, Milwaukee, WI). The images were evaluated subjectively for maximum resolution and definition at varying depths, overall contrast efficiency, and apparent image distortion.

A subjective evaluation reveals that although both imaging devices are capable of resolving the high contrast ⅛ in. level at the surface of the collimator (Fig. 2A and B, Fig. 3A and B), neither instrument is capable of resolving the ⅛ in. level utilizing the phantom described. It is also apparent that the two imaging devices are not equal in their capablity to resolve the larger disks at increasing depths. Although the image distortion demonstrated in Fig. 3 is apparent using standard bar and Smith orthogonal hole phantoms, it is also well defined using the phantom described. In addition, overall contrast efficiency is best perceived using the phantom described (Fig. 2C and D, Fig. 3C and D).

The phantom described simulates clinical conditions in the evaluation of total gamma camera system performance. It is large enough to evaluate the entire field-of-view of the large-field Anger systems. Our phantom shows the presence of linearity problems, image distortion, and resolution changes affected by contrast, depth and scatter. In addition, this phantom is valuable in generating optimal photographic contrast levels for lesion definition.

References

1. Graham MC, Anger RT: Instrumentation in scintimaging. In *Atlas of Scintimaging for Clinical Nuclear Medicine,* Wellman HN, ed, Cleveland, CRC Press, 1977, pp 1–12

2. Rollo FD: Techniques for evaluating imaging devices. *Appl Radiol:* 1–6, 1976

3. Kuhl DE: Physical principles of radionuclide scanning. In *Principles of Nuclear Medicine,* Wagner, HN, ed, Philadelphia, WB Saunders Co., 1968, pp 184-220