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THE RISE AND FALL OF THE RECTILINEAR SCANNER IN NUCLEAR MEDICINE

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The first use of localizing organs in Nuclear Medicine was with a probe. It consisted of a lead collimator for localization (not to be confused with the lead grid used to reject scatter in radiography / fluoroscopy). For quantification of thyroid uptake a standardized collimator was developed[1] in order to replicate the thyroid uptake around the world. To obviate the effects of the inverse square law a minimum distance from the face of the detector was required. When a GM probe was used there was no energy information available and the use of a filter was used to account for scatter (the lower energy scatter preferentially absorbed by the filter). The effects of scatter had to be accounted for and eventually were overcome by energy selection of the full energy response. The Compton scatter at lower energies must be rejected with the use of an energy discriminator commonly referred to as a pulse height analyzer (PHA). This is depicted in figure 1 where the photopeak of ^{99m}Tc is shown centered about 140keV while the Compton scatter and patient scatter of the spectrum is shown to the left at lower energies. The PHA window selecting only events from the photopeak to be included in the study is depicted by the vertical lines at 126 and 154 keV which constitutes a typical 20% window.
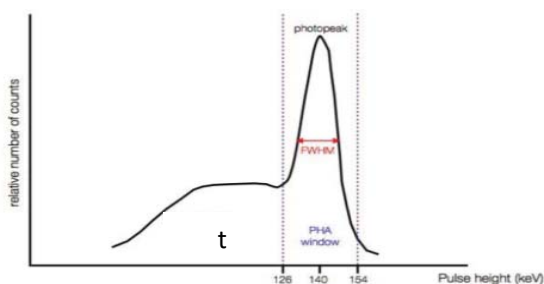
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Figure 1. A stylized energy spectrum of a typical ^{99m}Tc isotope including patient scatter which must be removed and was done by the PHA in nuclear medicine.

Probes were also used to give the time varying activity in organs, both for diagnostic work and importantly for radiation dose estimates. The use of two probes was very common to produce renograms of the kidneys.

A need existed for better definition of the organ. In order to address this need, the rectilinear scanner was invented by Benedict Cassen. A picture of this scanner is shown in figure 2.[2]

Figure 2. A photograph of the original Cassen rectilinear scanner shown on the right above the patient.

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It required a significant change in the collimator to obtain better definition of the organ. The detector/collimator was transported over the patient to set up the field of view(FOV).. The collimator consisted of multiple conical shaped holes, all whose apexes were at a fixed point. An image of the assembled head of a scanner is shown in figure 3.
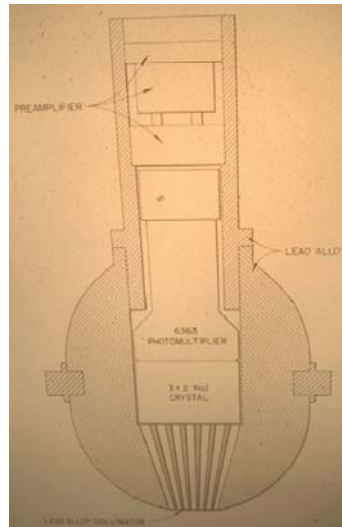

Figure 3. A typical early detector/collimator assembly for a rectilinear scanner for nuclear imaging. Included is the focused collimator on the front of a then typical 3" diameter crystal. Final designs settled on 5" diameter crystals prior to end of scanner development.

There were multiple conical holes separated by sufficient septal thickness of lead to preserve the integrity of the hole at the energy of the gamma ray used. The side of the holes need not have been circular, but the conical nature is required. This gives rise to a tomographic field of view with the spatial resolution being best at a distance from the face of the collimator. Therefore, a focal zone existed where the spatial resolution was satisfactory.
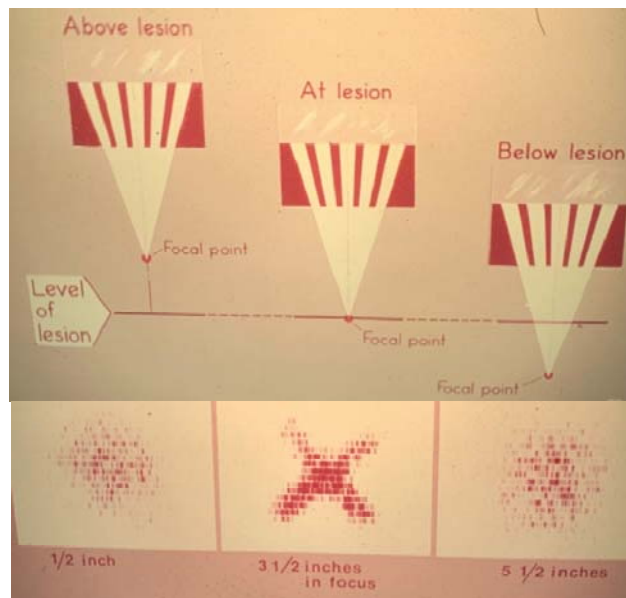

Figure 4. Top image depicts the focal or tomographic plane for a focused collimator. While the bottom image is an acquisition made of a planar radiation target in the shape of an X on the left above the plane of focus, middle in the plane and right below the plane. You can easily observe the natural blurring above and below this plane on the left and right while superior resolution at the plane of focus.

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Thus "if the image was clear, it was at the focal zone" as shown in figure 4. This is to be contrasted with the scintillation camera, where the holes are parallel and aren't conically placed. The spatial resolution of the camera using a parallel hole collimator is best at the surface and getting worse as you go away from the surface in contrast to the focal zone of the scanner's collimator. This very important point is shown in figure 5. Notice also included in figure 5 is a method employed in some instances to elongate the tomographic plane of view of the scanner by making the edges of the collimator holes as parabolic rather than straight while maintaining the lateral resolution at the plane of focus.
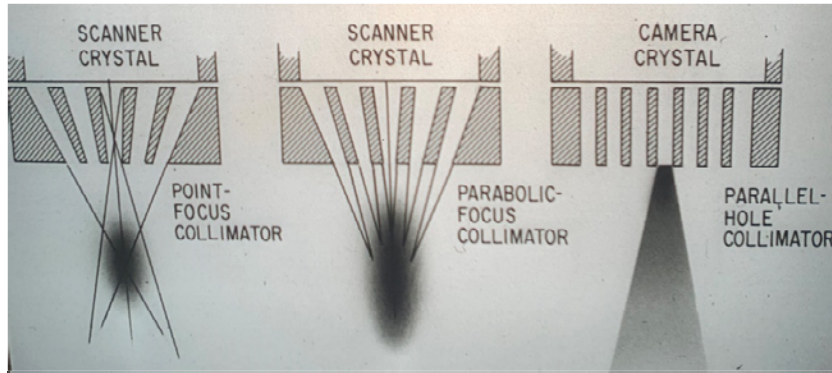

Figure 5. A comparison on the left and middle of a focused and parabolically focused collimators(used on scanners) to that of a parallel hole collimator(used on cameras). The focal or tomographic plane of acquisition was determined by the distance of the scanner detector from the patient while the cameras best resolution was at the face of the camera collimator.

The second very important point is that the field of view (FOV) of a scanner is set up by the scanning motion and can be quite large. The image was produced by moving the probe over the patient in a rectilinear fashion with the use of two motors. The motion in one direction across the patient was smooth while the motion in the other was a step, giving the rectilinear patterns. Initial scanners had a only one detector above the table while later scanners had detectors above (anterior projection) and below (posterior projection). This is depicted in figure 6.
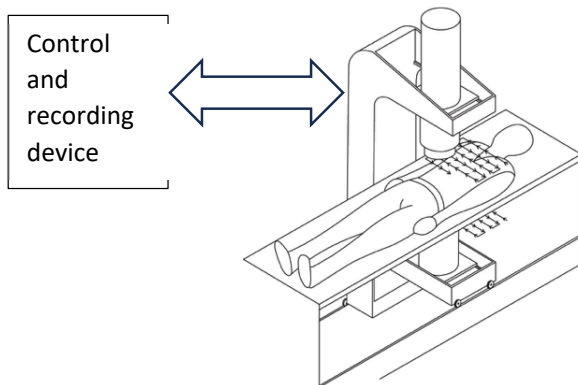

Figure 6. A generalized representation of a rectilinear scanner with a detector shown on the top and bottom of the patient. The detectors as shown would move laterally across the patient and then step to the next line and so on until the entire desired FOV would be covered. The detectors would be connected either mechanically in early devices to a dot making device recording either to paper or film and later electrotonically to some photoscanning film recording device.

In comparison, the FOV for the camera is setup by the area of the detector/camera head. You must keep this in mind when you consider the demise of the scanner. The essential tradeoff of spatial resolution of the collimator versus sensitivity must be made with both the scanner and the camera. The comparison of the two systems roughly depends on comparison of the sensitivity of the active area of the detector. The detector/collimator of the scanner was attached to the readout (either physically in the beginning and electronically in later vintages). In the case of the early scanners this was a physical arm connecting the detector to a device that could tap on a recording medium such as paper which by its nature gave a 1:1

representation of the organ being imaged, which gave a very useful additional property of the scanner in that your could directly determine the organ size as shown in figure 7.
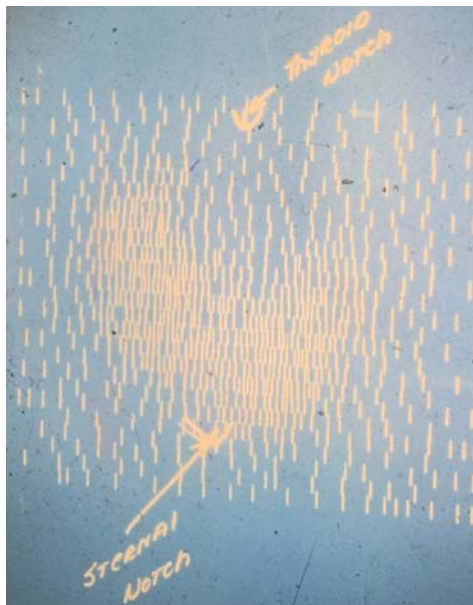

Figure 7. a paper recording of a thyroid using a mechanically coupled recording device to the scanner detector, where dots represent gamma events detected within the selected energy window and measurements could be made with a ruler directly.

The sensitivity of the scanner for single detector scanners was limited by the diameter of the crystal (5" diameter x 2" thick NaI(Tl)). The 5" diameter was limited by the focal zone in distance from the collimator and vertical size of the focal zone. The 2" thickness was sufficient for energies of ^{131}I and even higher energies. In contrast, the modern scintillation camera's 3/8" thick crystals are sensitive at low energies, such as that emitted by ^{99m}Tc, and very insensitive to higher to medium energies (<400 keV) e.g. ^{131}I. A comparison of this whole-body images from a scanner and a camera illustrates these points in figure 8.
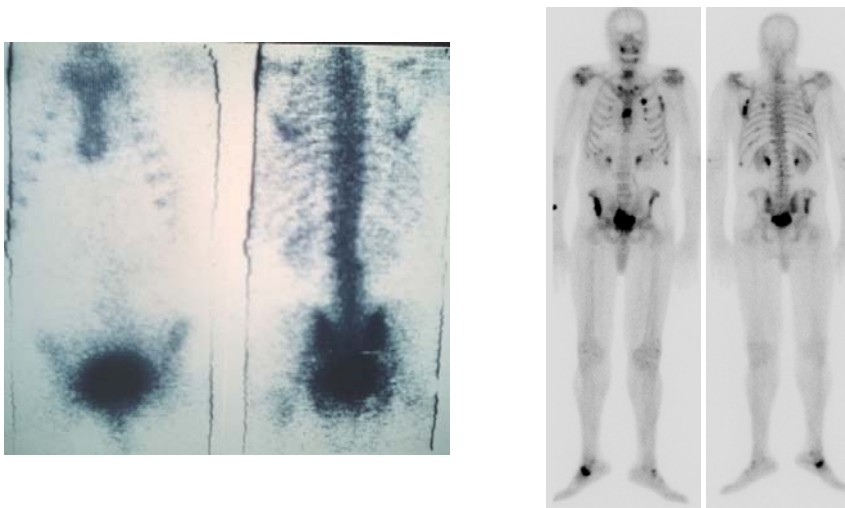

Figure 8. A whole-body exam both AP/PA for a scanner on the left and camera on the left.

While it is obvious which camera image is AP and PA, the scanner images are dependent on the operator putting the focal zone in the correct place for the scanner assuming this is physically possible given the mechanical nature of the detector and table. As discussed earlier, the rejection of scatter is essential in nuclear imaging. For the scanner, the PHA is fed into a count

rate meter which along with the speed of the motor and output size produces the image. Thus, the image is a representation of counts/area in the image. The camera output is also in counts/area. Figure 9 shows the significant effect of this statistical nature of the image upon perception. As depicted on the right even with a sufficiently large diameter object the observer is contrast limited by the low count density.
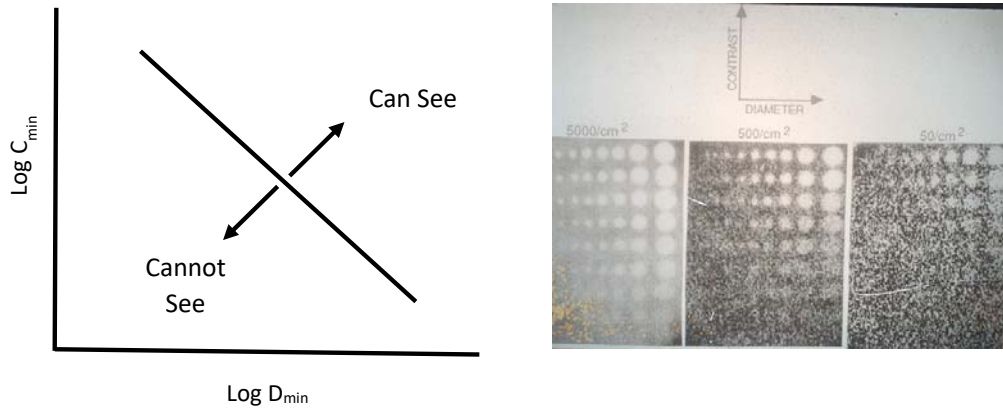

Figure 9. Depicts the limiting nature of statistics on observable contrast versus resolution in nuclear medicine. Depicted on the right is an old slide of a test phantom used in nuclear medicine showing how the observer becomes limited to identify larger diameters based on lower counts densities. This is graphically explained on the left.

This perception is most affected at low contrast. The advent of the scintillation camera[3] spelled the doom of the scanner. The area of the scintillation camera not only provided a means by which to view a large organ "in total" at the same times, allowing dynamic studies, but also increased the sensitivity. This efficiency gain was dependent on the use of ^{99m}Tc. The adoption of kits for ^{99m}Tc happened at the about the same time as the use of the camera, and worked in the favor of the camera. A useful timeline of these events is included at the end of this paper. The bone scan held on for a long time until the advent of the even larger area FOV cameras. While there were intermediate size cameras before the cameras of today, but as shown in figure 10 the intermediate cameras were of sufficient size to "stitch" together an adequate set of spot views which with the superior sensitivity of the camera was deemed better than the scanner.
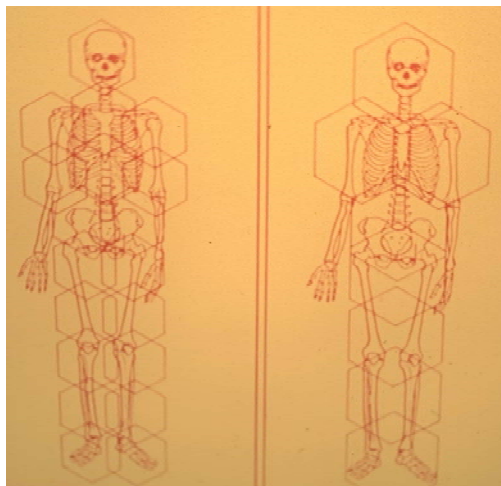

Figure 10. The whole body scan using older cameras by doing multiple spot images of the body on the left a smaller camera than on the more practical right side is depicted.

There was an attempt to use the tomographic imaging in a single direction with a rectangular detector, but the camera still won the day. When a moving table and the multi format images were introduced the increase in sensitivity was maintained and the "whole" bone scan was visible in one image for the camera. Thus, even the bone scan disappeared with the expanded

availability of new ^{99m}Tc radiopharmaceuticals. The use of the rectilinear scanner ended and the Anger scintillation camera took its place.

Keep in mind the significant differences

1) The area of the detector of the camera won the day
2) The spatial resolution of the systems are different.
    1. Camera is best at the surface and gets worse as you move away from the surface. The patient must be as close as possible to the collimator
    2. The scanner has a focal zone of acceptable resolution at a distance from the surface of the collimator. This must be placed at what you want to see.
3) The FOV of the camera is set by the area of the detector. A moving table must be used to obtain whole body contiguous images. The FOV of the scanner is set by the scanning motion and can be quite large, e.g., whole body.
4) Scanners were more efficient with ^{131}I but the use of ^{99m}Tc obviated this factor.
5) The acquisition time for cameras was far superior than scanners especially as camera sizes increased.

Historical Timeline [4]

1950 K.R. Crispell and John P. Storaasli used iodine-131 labeled human serum albumin (RISA) for imaging the blood pool within the heart.

1950 Abbott Laboratories sold the first commercial radiopharmaceutical, iodine-131 human serum albumin (RISA).

1951 The U.S. Food and Drug Administration (FDA) approved sodium iodide 1-131 for use with thyroid patients. It was the first FDA-approved radiopharmaceutical.

1951 Benedict Cassen, Lawrence Curtis, Clifton Reed and Raymond Libby automated a scintillation detector to "scan" the distribution of radioiodine within the thyroid gland.

1954 David Kuhl invented a photorecording system for radionuclide scanning. This development moved nuclear medicine further in the direction of radiology.

1955 George V. Taplin used iodine-131 labeled rose bengal to image the liver. He also used radioiodinated hippuran to measure kidney function with scintillation detectors.

1957 W.D. Tucker's group at the Brookhaven National Laboratory invented the iodine-132 and technetium-99m generator, making these short-lived radionuclides available at distant sites from the production of the parent radionuclides.

1958 Hal Anger invented the "scintillation camera," an imaging device that made it possible to conduct dynamic studies.

1959 Picker X-Ray Company delivered the first 3-inch rectilinear scanner.

1960 Louis G. Stang, Jr., and Powell (Jim) Richards advertised technetium-99m and other generators for sale by Brookhaven National Laboratory. Technetium-99m had not yet been used in nuclear medicine.

1962 David Kuhl introduced emission reconstruction tomography. This method later became known as SPECT and PET. It was extended in radiology to transmission X-ray scanning, known as CT.

1962 John Kuranz, Nuclear Chicago, delivered the first commercial Anger camera to William Myers at Ohio State University.

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1963 The FDA exempted the "new drug" requirements for radiopharmaceuticals regulated by the Atomic Energy Commission.

1963 Henry Wagner first used radiolabeled albumin aggregates for imaging lung perfusion in normal persons and patients with pulmonary embolism.

1964 The FDA exempted the "new drug" requirements for radiopharmaceuticals regulated by the Atomic Energy Commission.

1964 Paul Harper and Katherine Lathrup developed radiotracers labeled with Tc-99m for the study of brain, thyroid and liver.

1970 W. Eckelman and P. Richards developed Tc-99m "instant kit" radiopharmaceuticals. The first one was Tc-99m-DTPA.

1970 The FDA announced that it would gradually withdraw the exemption granted to radiopharmaceuticals and start regulating them as drugs. The change would be completed by January 20, 1977.

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