Clinical Applications of
Low-Energy High Transmission Collimator[4]

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The practice of medicine calls for continued efforts to improve its methods and materials. For example, in radiography at least three of the primary targets for improvement are: 1) sharp radiographic detail attainable by small focal spots and fine grain film emulsions, 2) contrast media for more selective organ study with reduced side effects to the patient and 3) further reduction of body radiation dose. These targets are paralled by those recognized in the practice of nuclear medicine. Great strides are being made toward these goals in both fields.

In recent years more and more attention has been directed to target organ radiation dose as well as whole body radiation dose. From this a trend toward clinical utilization of low-energy emitting isotopes began. ^{125}I was one of the early clinically-useful radionuclides having energies in the low range as compared to the midrange energies of ^{131}I, ^{198}gold, ^{51}chromium and ^{203}mercury. ^{203}Mercury in the form of chlormerodrin was proposed by Bender and Blau (1) as a tracer for brain tumor localization and for renal scanning. From the physical standpoint this isotope was ideal for scanning for it had a half-life of 45 days and a midrange energy emission (279 kev) that suited commercial scanning collimators. Since the target organ for mercury was the kidney, the midrange energies

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of the emissions and the relatively long half-life, so ideal for scanning, yielded a significantly high dose of irradiation to the kidney despite prior renal tubular loading with intramuscular injection of stable Mercuhydrin. The relatively low energy [197]mercury (in chlormerodrin) was then proposed by Sodee (2) and subsequent clinical trials reported by Overton et al (3) showed quite comparable results for brain scanning when chlormerodrin [197]mercury and chlormerodrin [203]mercury were used. [131]I human serum albumin served a useful function in myocardial scanning by providing a means of differentiating myocardial dilatation from pericardial effusion (4). This procedure was based on the labeling of the blood pool in the cardiac chambers, but gave no true scan image of the cardiac mass. [86]Rubidium was found by Carr (5) to have a relatively high affinity for myocardial muscle. However, the emissions of [86]rubidium (1.06 mev) caused poor performance with the clinical scanners commercially available, because this energy could not be adequately collimated with standard equipment. Cesium acetate ([131]Cs) was proposed by Carr (5) as a suitable myocardial tracer when he was able to discern a satisfactory separation between the count rate of peripheral blood as opposed to the count rate of the myocardium. He also found a higher heart/liver ratio of cesium activity than was observed using [86]rubidium. The preliminary reports of his work at the Ninth and Tenth Annual Meeting of The Society of Nuclear Medicine were encouraging and led to our initiating a program of myocardial scanning of animals and humans using our commercially-available scanner.[1] In our early efforts to scan the low-energy emitters we employed the 19 and the 37 hole focused collimators. We encountered extreme difficulty in separating the peaks due to the [131]cesium[2] K x-rays from the electronic noise of the photomultiplier tube, preamplifier, and spectrometer of our system. We compared our detector system with a three-inch Harshaw Integral-line detector which had a thin crystal can face and an improved photomultiplier system. In this comparison we observed a definite disparity of count rate between the two detectors; the integral-line detector showed at least a 50 percent increase in count rate. Incorporating the detector into our clinical system made resolution of the low energy peaks possible and yielded a definite increase in count rate for the low-energy as well as the midrange energy emitters. The original detector, it must be noted, had served perfectly adequately in the midrange energy isotope work, the purpose for which the scanner was developed many years ago. The same may be said for the commercially-available 19, 37 and 61 hole collimators with regard to midrange energy emitters, but the low energy emissions of [131]cesium, [197]mercury and ^{125}I do not penetrate well the usually thick aluminum scintillation crystal can, nor are the massive septa of the standard collimators necessary for scanning these low-energy emitters. Therefore, the low-energy, high transmission, 109 hole lead foil collimator prepared by Craig Harris et al at ORNL was employed for this study (6). Figure 1 reveals the scan image of a point source (^{125}I-loaded resin bead) scanned with the 19, 37, and 109 hole collimators. The source was placed at the focal distance for each

collimator and the scanning parameters were not varied nor was background suppression applied to the photo-record. It is obvious that the 19 hole collimator has a recordable sensitive volume that is at least one inch in diameter; the 37 hole collimator has a sensitive volume of approximately one-half inch diameter while the 109 hole collimator has a recordable sensitive volume of approximately one-quarter inch. On clinical application of this collimator, however, there appeared the ever-present scanner's dilemma, namely, the low count rate resulting from the increased degree of collimation. In view of the fact that low-energy emitters were applied in this study, the dose scale could be safely increased to provide a more adequate counting rate for the low-energy collimator without excessive irradiation hazard to the patient. Our initial investigation was directed toward myocardial scanning of the mongrel dog. The 19 hole collimator yielded a cardiac image having relatively even activity distribution with little separation of the myocardial and hepatic shadows (Fig. 2). The 109 hole collimator, on the other hand, showed a more irregular pattern of activity overall (due to low count rate) but did outline a great vessel, presumably the aorta on the right lateral scan
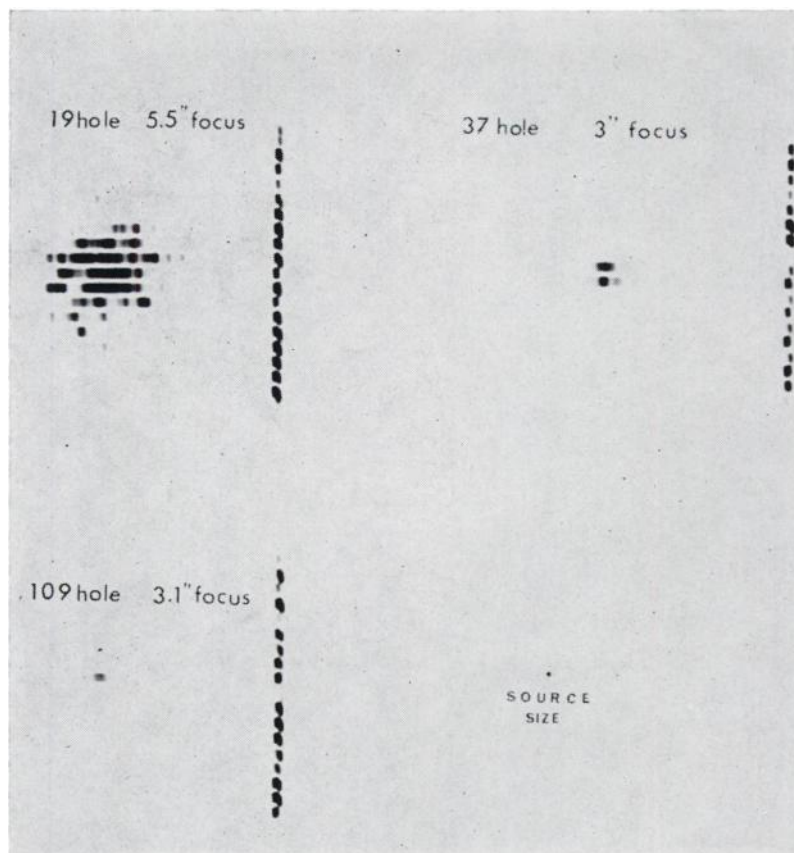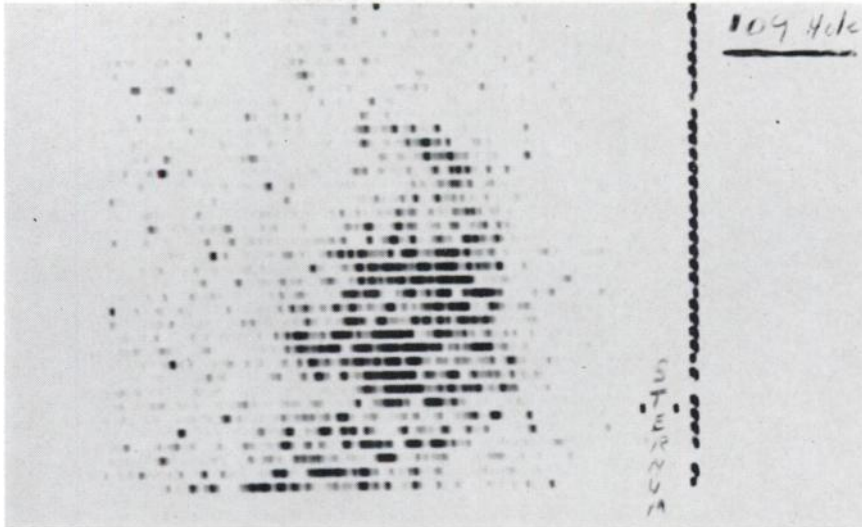

1. Scans of point source of ^{125}I were made at focal length of each collimator using no background suppression and no alteration of scanning parameters.
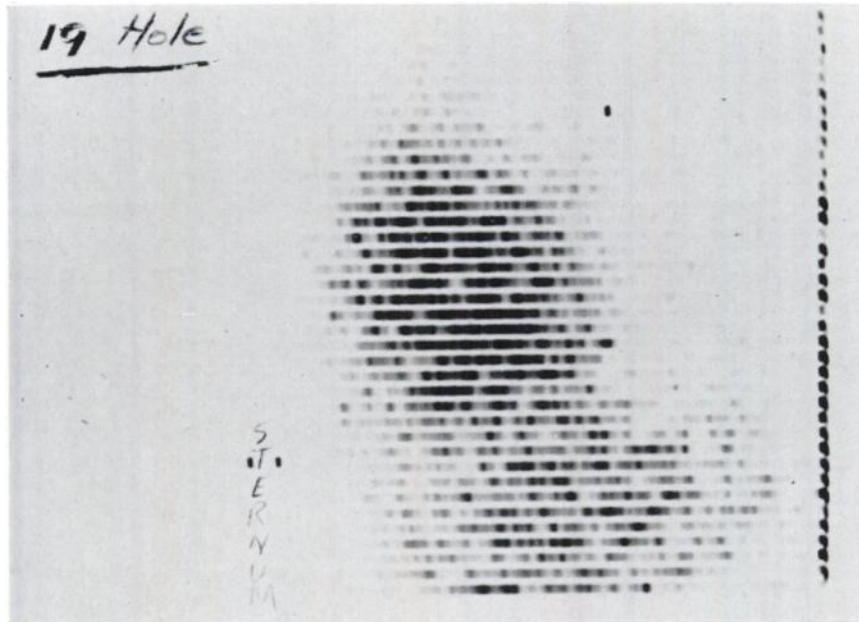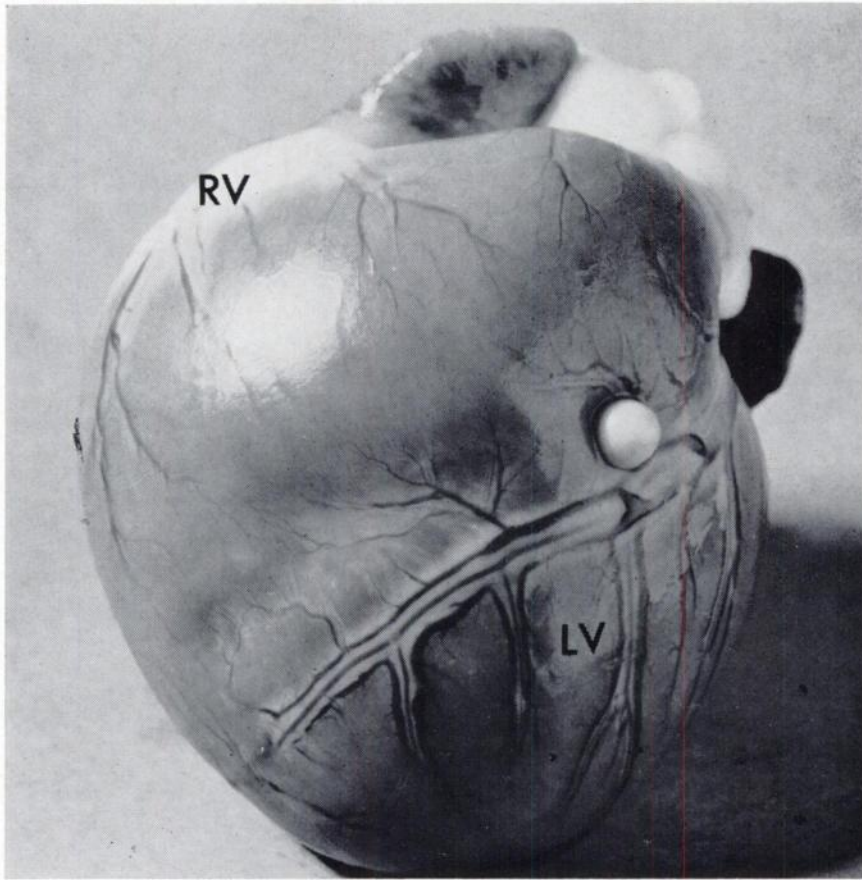
2. In vivo myocardial scan of dog following injection of 60 μc ^{131}Cs per kilogram for total dose of 1800 μc (19 hole standard collimator).
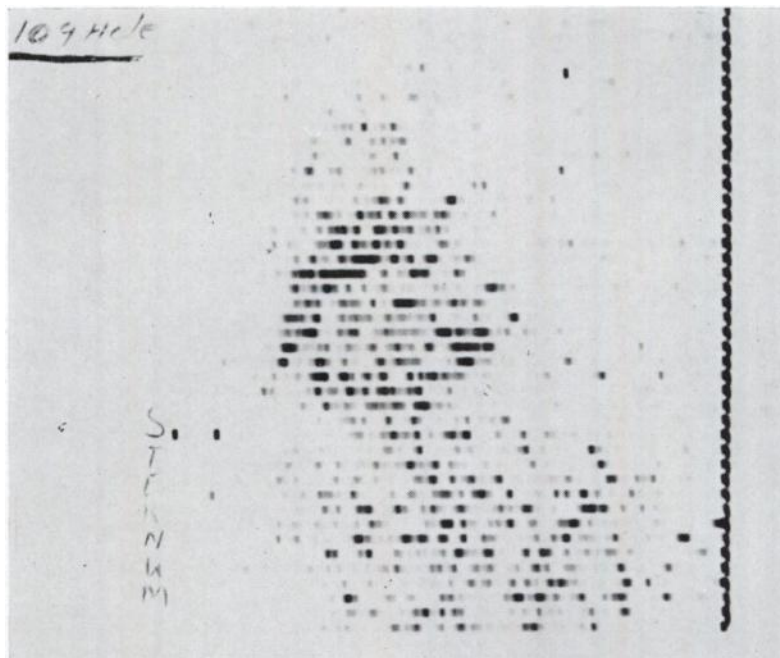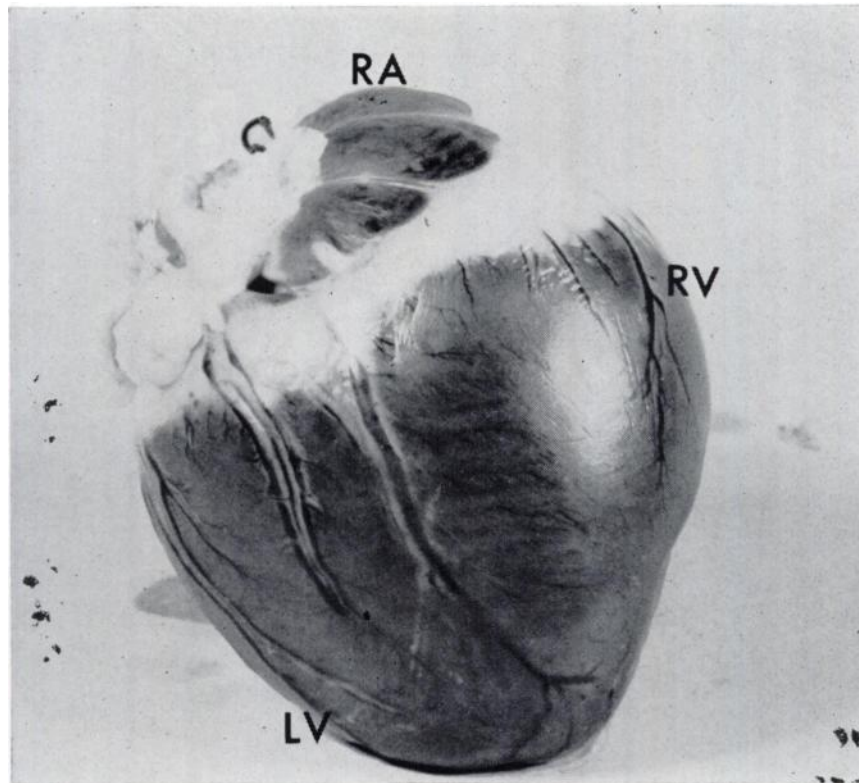

3. Left lateral dog myocardial scan utilizing low-energy collimator (109 hole).
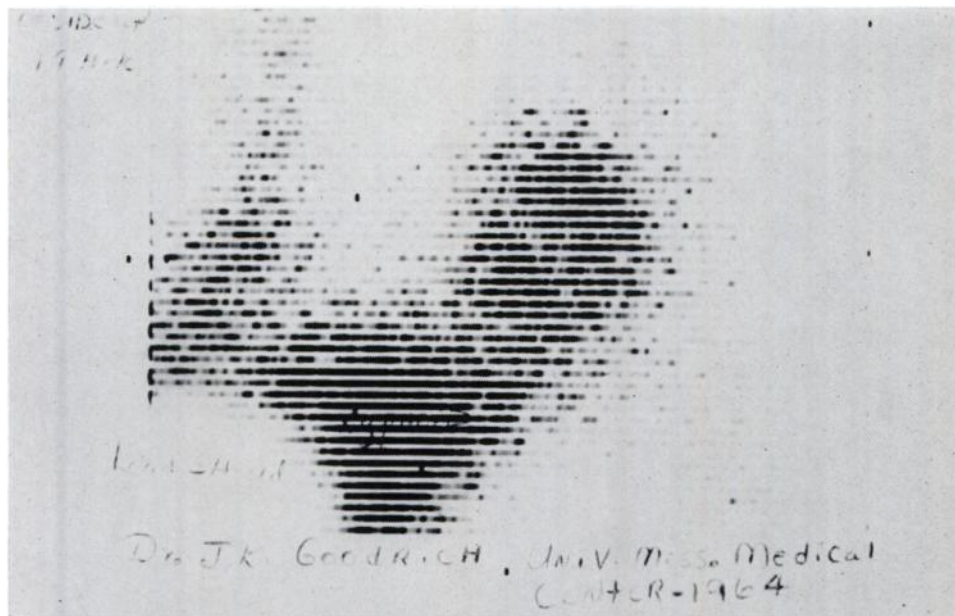
4. Right lateral dog myocardial scan utilizing low-energy collimator (109 hole).

5. Left view of heart removed from animal of scan shown in Figs. 3, 4.

6. Right view of heart in Fig. 5.

7. Left oblique scan of human myocardium using 19 hole collimator following injection of 2 mc of [131]cesium acetate.

(Fig. 3, 4). An area of diminished activity in the midportion of the cardiac image on the left lateral scan (Fig. 3) was viewed with suspicion when no definite counterpart on the opposite scan was found. The animal was killed, and the heart was removed but no definite evidence of myocardial ischemia or disease was found (Figs. 5, 6). Studies of sections of the myocardium in the well counter did not provide convincing evidence that any one area contained significantly higher concentration as compared with any other area of the myocardium. Repeated myocardial scans in the dog provided similar findings, although no other suspicious defects in the myocardial activity were shown. Myocardial scanning was then carried to human subjects, and here a severe limitation of the tracer was found. The low-energy emissions of ^{131}cesium would not penetrate the sternum and were severely attenuated by the costochondral structures. Scanning a normal myocardium *in vivo*, using the 19 hole collimator, two hours following injection of 2000 μc of cesium acetate ^{131}Cs revealed an image of the myocardium divided vertically by the sternum (Fig. 7). The distribution of activity appeared relatively even throughout the two portions of the myocardium, right and left of the sternum, but when the subject was scanned using the 109 hole low-energy collimator (Fig. 8) we were able to discern a suspicious defect in the left heart concentration, which, on reviewing the scan with the 19 hole collimator, could be detected faintly. This undoubtedly represents the course of one of the anterior ribs and its costal cartilage. This led us to believe that such defects might be misconstrued as actual failures of concentration of the tracer in the myocardium, as one might expect in a myocardial infarction. Furthermore, if the myocardial activity can be in effect hidden by the sternum and ribs, then a relatively small portion of the heart is actually observed directly by the scanning technique. We noted that the portion of the myocardium lying to the right of the sternum had a significantly lower concentration than that portion of the heart lying to the left of the sternum. This we attribute to the mass of the heart muscle lying to the left of the sternum and to the relative proximity of this portion of the heart to the anterior chest wall, as compared to the relative separation from the chest wall and the significantly lesser mass of the myocardium lying to the right of the sternum. A scan of a known myocardial infarction provided gratifying results despite the feared attenuation defects of sternum and rib cage. Scanning the heart of this known myocardial infarct patient using the 19 hole collimator (Fig. 9) revealed a definite defect in the lower margin of activity in the portion of the heart lying to the left of the sternum. The maximum counting rate of the heart using the 19 hole collimator was 8,000 counts per minute. A repeat scan using the 109 hole collimator yielded a maximum count rate of 1000 counts per minute (Fig. 10) and here again the defect in the activity concentration along the lower margin of the myocardium was clearly delineated. When the patient was turned in a left anterior oblique projection we found the defect moved slightly nearer the sternum as one might expect in an anterior-lying lesion (Fig. 11). The dimensions of the defect shown by scanning with the 109 hole collimator are not significantly different from those recorded using the 19 hole collimator.

Having satisfied our requirement that a myocardial infarction of a sizable volume could be detected using the 19 hole collimator and shown to at least
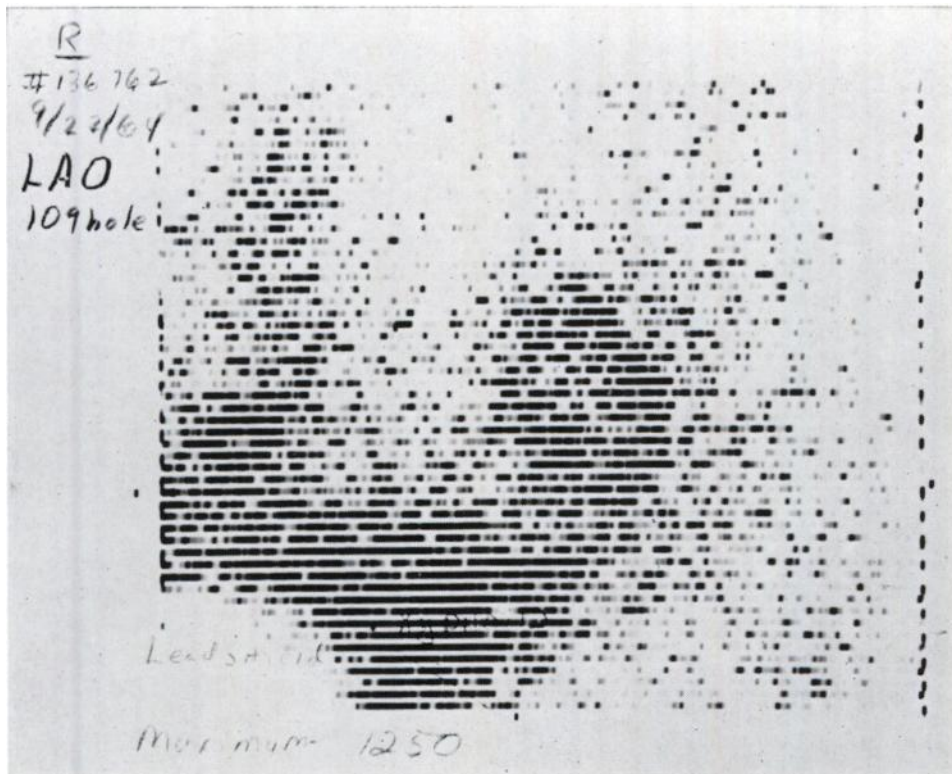
8. Repeat scan of subject in Fig. 7 using 109 hole collimator. Low count rate accounts for salt and pepper pattern and lack of contract. Note defect in activity distribution in heart to left of sternum which is caused by overlying rib.
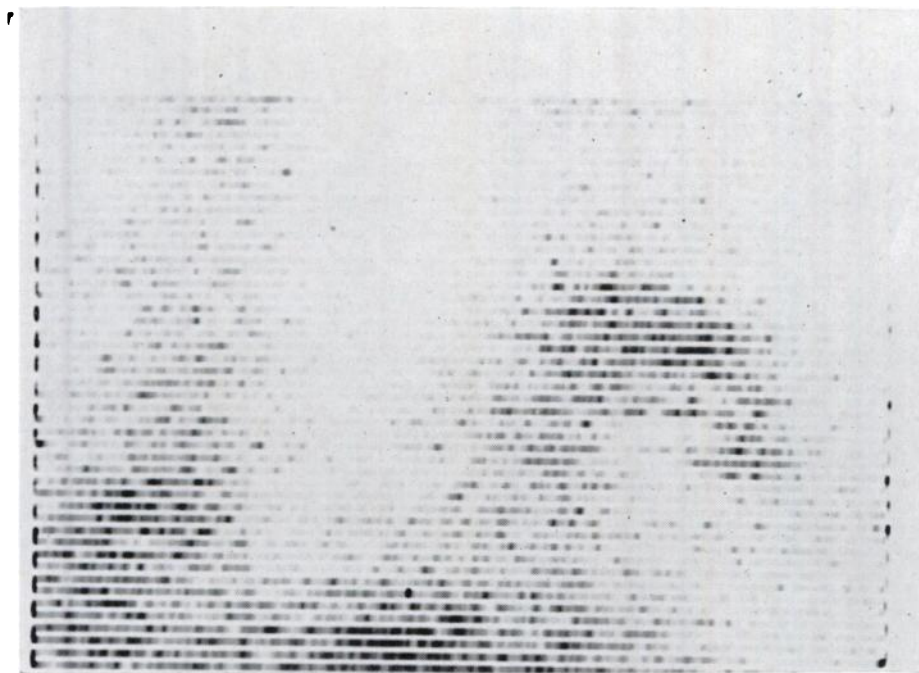

9. AP myocardial scan using 19 hole collimator. Sixty-year-old patient having known myocardial infarction was the subject. Note circular defect in lower margin of cardiac scan image to left of sternum at site of infarction.
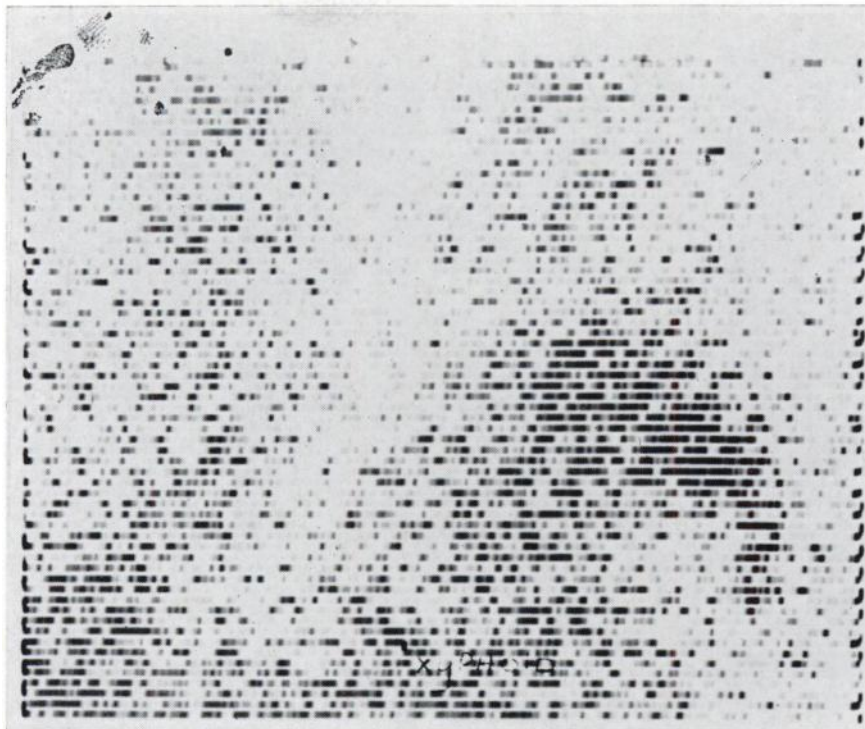
10. Repeat of Fig. 9 scan using 109 hole collimator. Note myocardial infarction defect visible without notable change in dimension.
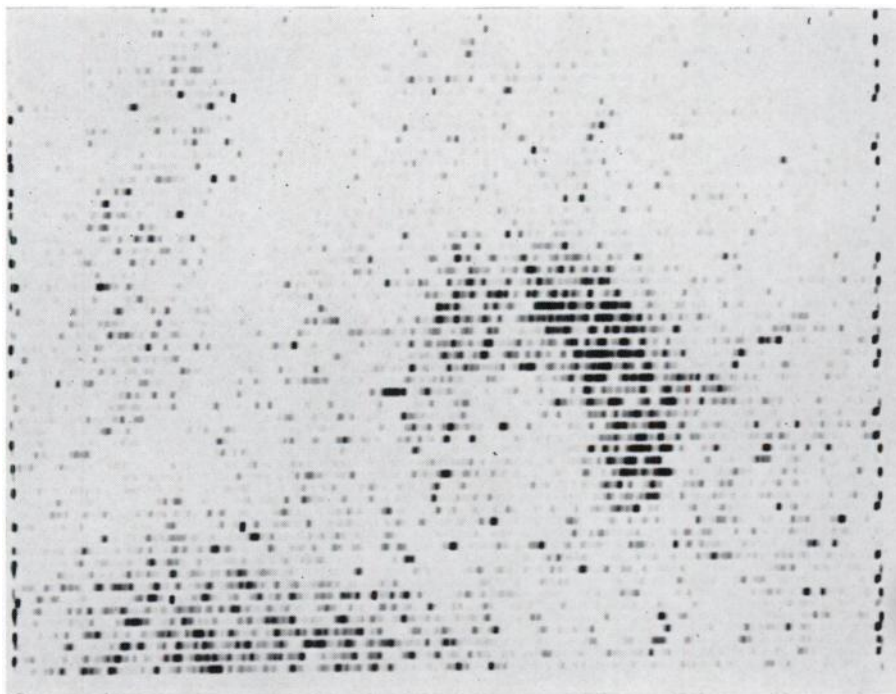
11. Left oblique scan using 109 hole collimator. Note obvious disparity of count rate between heart to right and left of sternum.
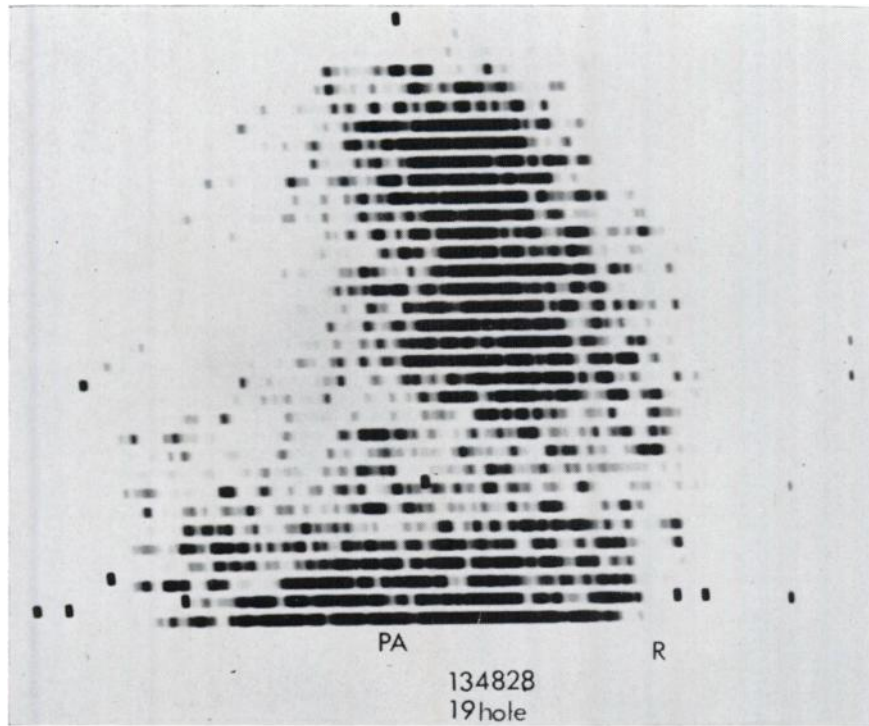
12. PA brain scan using 19 hole standard collimator. Patient received 700 μc of ^{197}-mercury chlormerodrin. Clinical diagnosis of CVA was made.
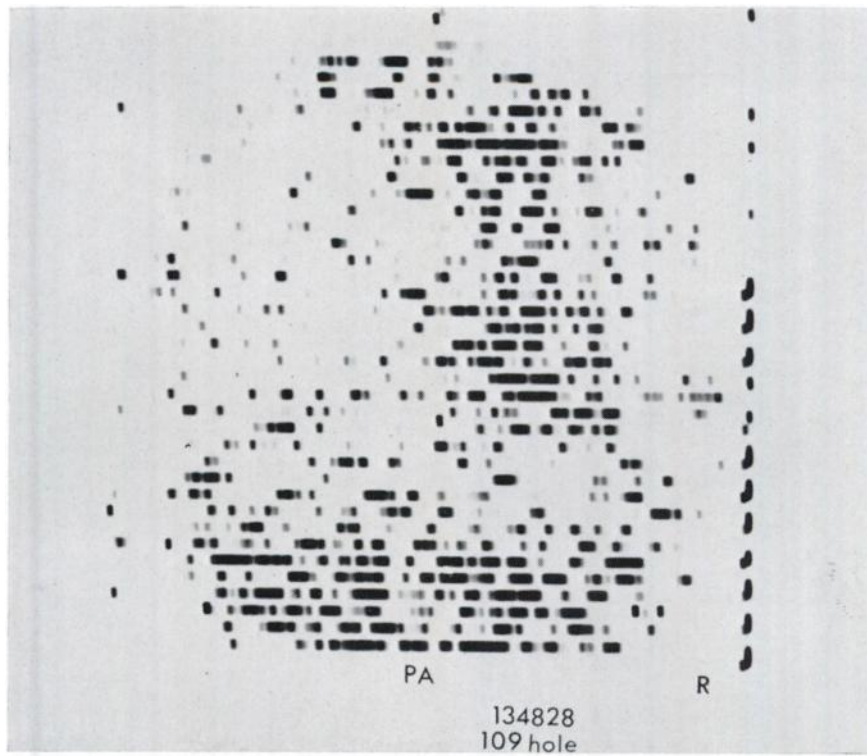
13. Repeat scan of patient in Fig. 12 using low-energy collimator (109 hole). The lesion is clearly visible through recorded pattern of activity is altered by reduced count rate.

equal satisfaction utilizing the 109 hole collimator, we then approached the prob-
lem of scanning of the brain using the low-energy emitter, [197]mercury in chlorme-
rodrin[1], in doses ranging from 300 to 1000 μc. The 19 hole collimator recorded
clearly the intracranial lesions having high activity concentration, such as that
in (Fig. 12). The 109 hole collimator recorded the lesion clearly, but a severe
reduction of count rate was seen (Fig. 13). We received some encouragement
however, in the usefulness of the fine resolution of the 109 hole collimator when
we demonstrated the confluence of sinuses on a PA scan (Fig. 14), a finding here-
tofore not obtained on scanning with the 19 hole collimator. This delineation of
a normal vascular structure supports our therory that greater detail may be ob-
tained with the 109 hole collimator, yet the chief limiting factor in producing a
brain scan image-which is comparable with the images obtained using chlor-
merodrin [203]mercury and the 19 hole collimator is the severe reduction of count
rate. The photoscan image obtained in low-count-rate situations has a "salt and

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[1] [197]Mercury chlormerodrin supplied by Medotopes Division, E. R. Squibb and Sons.

PA   RT
134216
109 hole
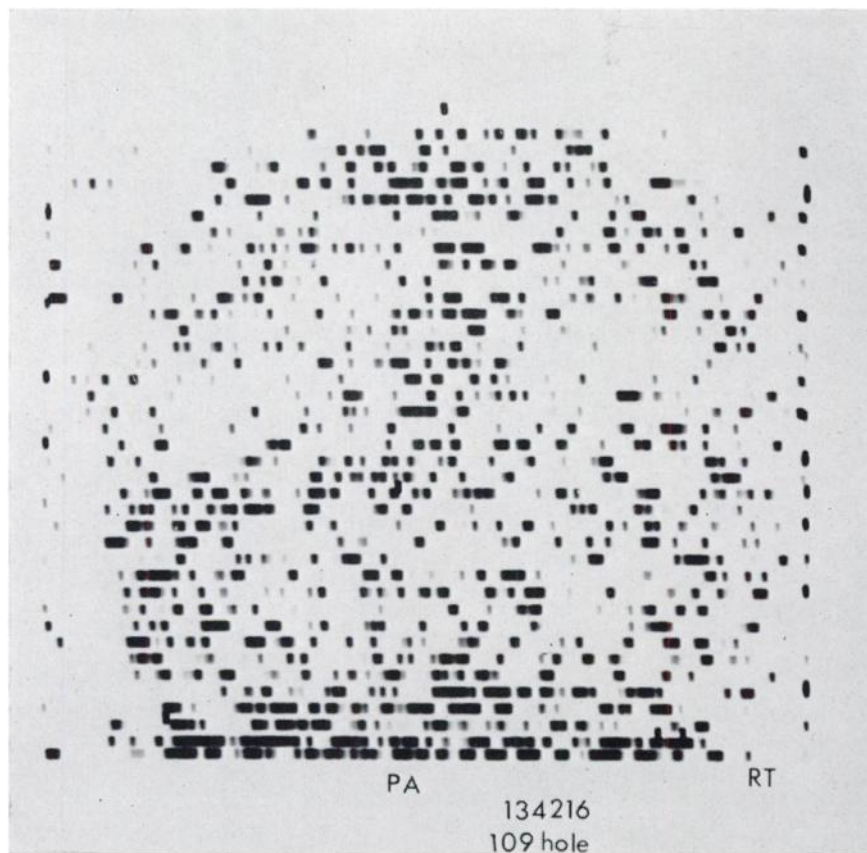
14. PA brain scan of patient having no tumor or vascular abnormality. The 109 hole
collimator delineates activity in the confluence of sinuses.

pepper" pattern which is difficult to interpret. A basic concept to the photographer is the need for photons of light to produce a photographic image. When sufficient photons of light are present, then greater detail of the image can be obtained by narrowing the aperture of the camera lens. The optimal lens aperture setting varies according to the light value (photons available) and therefore no one lens setting will suffice for a wide range of photographic scenes. The same is true in collimator design for use in scanning the low-energy emitters. Utility of the low-energy emitter in the clinical setting depends therefore on the development of scintillation detectors and collimators capable of operation in the low-energy range. We propose that at this point the clinician is not faced with a requirement of complete revamping of his scanning equipment. Instead the application of appropriate low-energy collimators to standard scanning detectors may provide the necessary intermediate step between midrange energy scanning and combination scanning of both low and midrange energy emitters.

The art and science of body organ scanning have reached the point where both isotopes and detector-collimator combinations must be tailored to fit a specific diagnostic chore. In time, this will require a relatively wide selection of collimators, but in the end will provide us with results that approach the diagnostic efficacy of a good radiograph or contrast study of an internal organ.

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