(1) Multihole collimator

	Λ	B	C	
Nominal maximum gamma-raye nergy	0.20	0.36	0.41	MeV
Material	W	Pb	Pb	
Hole length	2.5	5.6	6.6	cm
Hole diameter	0.31	0.49	0.82	cm
Septum thickness	0.08	0.23	0.4	cm
Number of holes	4681	1573	559	
Λrea of view (in dia)	27	30	30	cm

(2) Fluorecent plate: zinc sulfate, 30 cm in dia
NaI (Tl) scintillator: 11″1/2 in dia, 1/2″ in thickness
(3) Objective optical lens: 4:40 mm F:0.95
(4) Multistage image tube: TOSHIBA M7064A
(5) Tandem lens: f:50 mm F:0.95 × 2
(6) TV camera: 1″ videcon with tandem lens
Video tape recorder and TV moniter are equiped.

Conclusion

1. Based on the ultrasensitive multistage image tube produced in the TOSHIBA Co., new sensitive gamma-camera was designed and evaluated.
2. This camera has its advantage in the freedom of design. Since original image is coupled optically on the onput surface of the image tube. Either sensitive crystal or inexpensive fluorescent plate can be used as the gamma-rays to light transfer material.
3. Phantom study revealed good resolution and sensitivity of this new vamera.
4. Animal and clinical application are reported.

Multistage Image Tube γ Camera Device and Radioisotope Angiography

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A new and simple γ-ray camera employing a couple of multistage image tubes as the intensifier with external NaI (Tl) crystal as the detector was introduced in 1967 by authors.

The 11½-inch-diameter crystal was coupled optically to the specially developed ultrasensitive image intensifier tube, Toshiba multistage image tube M-7064-A. This tube was originally designed for use as a dark image intensifier which naked eye scarcely percepts. When a γ-ray interacts with the crystal, light is emitted and is projected on the input end of the tube through the optical lens system. This light in turn causes photo-electrons from each part of the photocathode corresponding to the brightness of the input image. After intensifications by four multiplying dinodes these electrons are focussed onto an output screen with the light flux gain of 10^4 to 10^5.

This multiplying process is repeated by the additional second tube with resultant gain factor of 10^8 to 10^{10}. Thus each γ-ray that interacts with the crystal produces several thousand photons at the output screen, which is recorded on videotape recorder for further analysis.

The device is used mainly for the 140 KeV γ-rays from ^{99m}Tc. A multi-channel tungsten collimator with 4681 holes are employed to project an image of the subjects on the crystal.

Radioisotope angiography of the cases with control and various cardiopulmonary disorders were studied. After i.v. inj. of 10mCi of ^{99m}TcO4 saline solution the process of ^{99m}Tc bolus filling and escape from the heart chambers was recorded, processed later as the cummulative images of dilution process in one second intervals. Also with specially designed collimated light sensor, dilution processes of each heart chamber was obtained by replaying the tape. This method is very useful because the numerical data from a specific picture area recorded on the tape can be read out any time.

This camera has the advantages of simplicity, good sensitivity for both low and medium-energy γ-rays, and no dead time problems for large dose of radioisotope administered. Radioisotope angiography is one of the most promising application of this γ-camera.

A Method of Lung Profile Scanning

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The purpose of this report is to describe our method of lung profile scanning using ^{131}I-MAA and ^{133}Xe and our equipment with driving unit and speed-changing gears.

Two collimated counters with 2 inches crystal detector were placed behind the lungs of a subject, whose back was attached to the lucent plastic plate. The linear profile scanning was performed from the apex to the bottom of the lungs. Various speeds of scanning were obtained by combination of an induction motor and gears. Pre-setting markers set in the equipment determined the position of the detector to the lungs.

The apparent shift and decrease of peak values of the pulmonary blood distribution curves were examined, which resulted from combination of various scanning speeds and time constants of the rate meter using ^{131}I-MAA. When apparent shifts of curves were estimated by the variation of the upper to lower blood distribution ratio (U/L ratio), a real distribution curve was obtained in the following conditions, i.e., when time constant was 0.5 sec., the scanning speed was 20 mm/sec. or less, and the speed for time constant of 1 sec. was 5 mm/sec. or less. The decrease of peak value was not found in the following conditions, i.e., when time constant was 0.5 sec., the scanning speed was 30 mm/sec. or less, and for time constant of 1 sec. the speed was 10 mm/sec. or less, respectively. So that, optimal conditions of scanning speeds and time constants were determined to be a combination of 0.5 sec of time constant with 20 mm/sec. or slower of speed, and that of 1.0 sec. with 5 mm/sec. or slower.

The ratio of U/L in ^{133}Xe-pulmogram was slightly larger than that in ^{131}I-MAA pulmogram under the same conditions. This seems to be resulted from the alteration of the lung volume by breath-holding follows ^{133}Xe injection and the energetic difference between two isotopes. That is ^{133}Xe has low energy level (80 KeV) in comparison with that of ^{131}I-MAA (364 KeV). Since gamma-emitter (^{133}Xe) distributed in the deep and front part of the lungs was absorpted to considerable extent by the tissues, the rear detector could not detect sufficiently its emmission. In this respect, two detectors were confronted, one in the front and the other in the rear of the lung so that an efficient and even detection of the gamma-emmission was accomplished.

As the conclusion, our method presently described is useful in establishing clinical diagnosis and elucidation of pulmonary blood flow distribution.