Dynamic Performance of Scintillation Camera-Computer System and Correction for Counting Loss due to Resolving Time

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Dynamic image data obtained with Anger camera-computer system are often inaccurate due to finite resolving time of camera and computer sy stem. The resolving times of a delayline type scintillation camera and on-line computer system (TOSBAC 3400 Model 31 DAC system) were measured separately and overall correction methods for counting loss are established using the experimental equations. Resolving time of the camera was determined with fitting a curve of display countingrates vs. relative intensity of radioactive source with the following equation : N=RNo exp (-Tc No) where N is the display counting-rate, N. is the total detection event rate, R is ratio of display event to the total event and Tc is the apparent resolving time of the camera. Since the true counting-rate, Nt, is equal to R No, the above equation is rewritten as follows : N=Nt exp (−Tc. Nt/R). R is determined by pulse height analysis of z signal of the camera calculating the ratio of area under peak of interest to total area of whole spectrum. The resolving time of the computer system was determined by measuring the collected counting-rates (M) as a function of the display counting-rates (N). Experimental curves were well fitted by the following equation : N=M/(1-MT) where T is the system resolving time. Performance of computer system was also limited by the data transfer speed from core memory to magnetic disc. Overall counting-rate performance of camera-computer system is derived from above two experimental equations and future improvement is suggested.

Clinical Evaluation of Computer Data Correction System on Scintillation Camera

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Recently digital computer system is commonly used for scintillation camera data processing. Correction of scintillation camera field inequality is one of important problem. But, clinical evaluation of the correction system is not enough established.

The system we used consists of a on line digital computer with 8 K memory and delay line type scintillation camera with 20, 000 hole collimeter. Tc-99 m flood field souce of 30 cm diam, and many small phantoms (1—50 mm diam) were used for this study. The 11. 5-in-diam crystal of scintillation camera is currentry represented as a 64—64 matrix. A symple program was written to correct the inequalities on matrix element of computer.

Uncorrected scan was made with a sheet souce of Tc-99 m and 20,000 hole collimeter and then correction was made with the computer program using previously stored correction facters. The deviation of count on each was corrected from 11% to 1%. Then, many small phantoms of Tc-99m with same activity was set on many areas of camera and scan was made. ROI (region of interest) was selected enough to each small phantoms on 64—64 matrix. The deviation of each ROI was 19.2—12.2%. The correction was made with a correction facters obtained from sheet souce. However the deviation of ROI count 17.4%—9.8%. There is no comparison between sheet souce data and small phantom data. Quantitive studies on small phantom examination were made changing doses, form, size of phantoms, spectrometer window width and correction factors on time. The improvement on correction data was not obtained.

This study suggests that correction of scintillation camera field inequality by computer is unavailable on clinical studies such as renal scan, thyroid scan and so on.

Radioisotope Image Processing with a Digital Filter

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It is well known that two major cases make the radioisotope image obscure : (1) noise due to statistical fluctuations and (2) blurrig due to finite resolving power of a measuring system. In order to reduce the effect of the deteriorating cases or to enhance the information contained in image, a digital filter using the high speed Hadamard transform of RI image is presented.

The observed image is expressed by the convolution of true radioisotope distributions and the impuls response of instruments.

Then, $$g(x, y)=\iint_{-\infty}^{\infty} f(x', y')h(x-x', y-y')\,dx'dy'$$

and $g'(x, y)=g(x, y)+e(x, y)$.

Here, $g'(x, y)$ is the observed "digital RI image" under consideration, $f(x, y)$ is the true radioisotope distribution in a two dimmensional plane and $h(x, y)$ is the resolving power of system. For improveing the resolving power of the system, the Hadamard transform of the observed digital image is performed as follow : $[G'(u, v)]=[H(u, v)]\ [g'(u, v)]\ [H$