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EFFECTS OF POSITRON RANGE AND ANGULAR DEVIATION ON SPATIAL RESOLUTION IN POSITRON EMISSION COMPUTED TOMOGRAPHY.

N.Nohara and E.Tanaka. National Institute of Radiological Sciences, Chiba.

Increases in widths of line spread functions due to positron range and angular deviation were analytically evaluated. The line spread functions allow to make the evaluation free from slice thickness. The line spread functions in projection due to positron range were calculated from point spread functions experimentally obtained by Derenzo (Proc. 5th Intern. Conf. Positron Annihilation, Lake Yamanaka, Japan, April 1979). Increases in response widths of reconstructed image due to the positron range were computed for line sources of three radionuclides, C-11, Ga-68 and Rb-82, under the assumption that a coincident detector pair response is given by a Gaussian function. The line spread function of angular deviation was obtained from the data for water by Colombino et al.(Nuovo Cimento 38, 707, 1965). Widths of spread function in reconstructed image of a line source taking positron range and angular deviation into account were evaluated as a function of detector ring radius, for a coincident detector pair response of Gaussian function with 2 mm FWHM and 3.65 mm FWTM. For a 80-cm diameter detector ring, for instance, the FWHM increases from the coincident detector pair response of 2 mm to 3.16, 4.05 and 4.66 mm, and the FWTM increases from 3.65 mm to 5.97, 8.43 and 13.7 mm for C-11, Ga-68 and Rb-82, respectively.

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CORRECTION FOR BLURRING DUE TO FINITE POSITRON RANGE IN HIGH RESOLUTION POSITRON TOMOGRAPHY.

E.Tanaka and N.Nohara. National Institute of Radiological Sciences, Chiba.

The blurring due to finite range of positron can be compensated by modifying the convolution filter in image reconstruction. The modified filter g(i) is given by

$$g(i) * b(i) = g_0(i) * s(i)$$

where b(i) is the blurring function in projections, $g_0(i)$ a conventional (Shepp-Logan) filter, s(i) a corrected PSF which is assumed to be Gaussian, and * denotes a convolution. g(i) can be determined by an iterative method from known b(i) and s(i). The factor of increase in r.m.s. noise due to the correction is given by

$$F = [\sum_i g^2(i) / \sum_i \{g_0(i) * s(i)\}^2]^{1/2}.$$

Using Derenzo's experimental data (1976) for positron range, b(i), g(i) and F-values were obtained for various FWHM of s(i). Table 1 shows the F-values for two extreme cases: (A) a point source is placed at the center of a detector slice having a 4 mm width and (B) a line source is placed perpendicularly to the slice (shown in parentheses).

Table 1. Factors of noise amplification due to positron range correction

FWHM(mm)	C-11	Ga-68	Rb-82
1.0	2.06(2.11)	6.09(7.96)	- -
2.0	1.41(1.44)	3.68(4.66)	4.04(8.65)
3.0	1.17(1.20)	2.41(2.91)	3.48(7.12)

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INITIAL CHARACTERIZATION OF A NEW DESIGNED BGO DETECTOR.

S.Miura,S.Yamamoto,H.Iida, I.Kanno and K.Uemura.Research Institute for Brain and Blood Vessels, Akita.

The primary difficulty in designing a fine spatial resolution bismuth germanate (BGO) detector lies not in the detection properties of small BGO crystals, but in the isze of available photomultiplier tubes (PMTs).

Recently, a new desinged BGO detector have been developed in our group. The detector was constructed from eight small BGO crystals (3 mm x 24 mm x 30 mm depth), unique shaped lightguide and dual rectangular PMT, which has two independent segments in one glass envelop, each of which has a 12 mm x 24 mm entrance window (Hamamatsu, R1548). Throughout the lightguide, phontons from each BGO crystal were divided into two, the ratio of which is correlated in the relative position of the crystal and lead into two segments of the PMT. The position of each BGO crystal of the detector was discriminated using two analog outputs from the PMT.

We initially studied the characteristics of the detector to single γ-ray beam (511 KeV, less than 1 mm width) scan and obtained results as follows; 1) The peak position of the point spread moved linear with the scan. 2) The resolutions of the point spread were about 4.8 mm FWHM and 4.0 mm FWHM at the center and the both sides in the detector face, respectively. 3) The energy resolution was about 26 % FWHM.

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THE ESTIMATION OF THE STATISTICAL ERRORS IN THE MEASUREMENT OF rCBF USING THE STEADY STATE METHOD.

S.Miura,H.Iida,I.Kanno, M.Murakami,K.Uemura. Research Institute for Brain & Blood-Vessels, Akita.

The steady state method which can be used for measuring regional cerebral blood flow (rCBF), amplifies statistical errors, inherrent in the transmission and emission scan of the PET study, in the model used for calculating rCBF. Therefore, it is important to estimate the statistical errors in rCBF for the quantitative evaluation of rCBF. The method to estimate them was derived using the arterial concentration of ^{15}O(Ca) and the phantom study.

Method: The relation between Ca of the subject and the mean pixel count rate (Ct) of each region of the brain could be obtained from our clinical studies as follows:

$$Ct = k \cdot Ca \quad \ldots\ldots (1)$$

where k is a constant depend on each region of the brain. In the phantom studies, the statistical errors (ΔCt/Ct) in the PET image due to Ct could be obtained as follows:

$$\Delta Ct/Ct = g(Ct) \quad \ldots\ldots (2)$$

On the other hand, from the equation for steady state model, the statistical errors (Δf/f) in the due to Ct is:

$$\Delta f/f = (1 + f/\lambda)(\Delta Ct/Ct) \quad \ldots (3)$$

where λ is the decay constant of ^{15}O. Combining Eq. 1 and 2 into 3 gives:

$$\Delta f/f = (1 + f/\lambda) g(k \cdot Ca) \quad \ldots (4)$$

Eq. 4 means that the statistical errors in rCBF of each region of the brain results in being estimated from Ca of the subject.