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Based on a double-headed rotating uncollimated scintillation camera system, a positron imaging device was developed. After a rotating data acquisition in coincidence mode, 16 transverse section images are reconstructed by back projection. To obtain a uniform response, a limited angle reconstruction option is incorporated in this process. After correction for the system response by a three-dimensional deconvolution technique, the 16 transverse section images are stored on disk as a standard patient study for further analysis. The system can also be operated in a stationary mode. In this mode longitudinal tomographic images are obtained. Return to single photon scintigraphy is possible by remounting the collimators and by switching off the coincidence electronics.

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Most positron imaging devices are multidetector systems, built as a ring or hexagonal system. Each detector is measuring the annihilation radiation in coincidence with a number of opposing detectors. An extensive review on tomographic imaging has been given by Budinger et al. (*1*).

The use of a double-headed scintillation camera system, as already suggested by Anger in 1959 (*2*), has been explored by Muehllehner (*3,4*). Problematic was the demand on the linearity of the individual cameras in such a rotating system (Muehllehner G, private communication). With the general availability of rotating double-headed scintillation camera systems, in combination with the implementation of energy and linearity correction electronics into the camera electronics, a positron camera based on such a system is a useful and versatile instrument, since it can be used for all imaging modalities in nuclear medicine.

The system described in this paper consists of a double-headed rotating system with two ZLC cameras* operated in a coincidence mode. The implementation of the coincidence circuit was done without changes to the camera electronics, so return to single photon scintigraphy is possible by switching off the coincidence electronics and by remounting the collimators. After a rotating data acquisition, 16 transverse section images are reconstructed by back projection followed by a three-dimensional deconvolution in order to correct for the system response function. It is evident that no dynamic studies can be performed due to the time required for rotating the camera system. The system can also be used in a stationary mode in which longitudinal tomograms, parallel to the detectors are reconstructed by back projection (*5*).

SYSTEM DESCRIPTION

The Rota system consists of two ZLC cameras with 39 cm diam, 3/8-in.-thick crystals. Both cameras are mounted in a rotating gantry. For the conversion of the system into a positron camera the following hardware and software was implemented:

1. Coincidence electronics;
2. Data acquisition software which also controls the rotation of the gantry;
3. Back projection software which back projects the coincidence events into a user-definable three-dimensional volume;
4. Deconvolution software to correct for the system response function.

The coincidence electronics as developed for the stationary positron camera equipped with two large field of view (LFOV) cameras, A and B, could be used with only a minor modification due to timing differences between a LFOV and a ZLC camera (*5*). The coincidence electronics deliver four position signals (X_A, Y_A, X_B, Y_B) if a coincident 511 keV event has been detected. In case the coincidence unit is activated but no coincidence event is established within the time window, a

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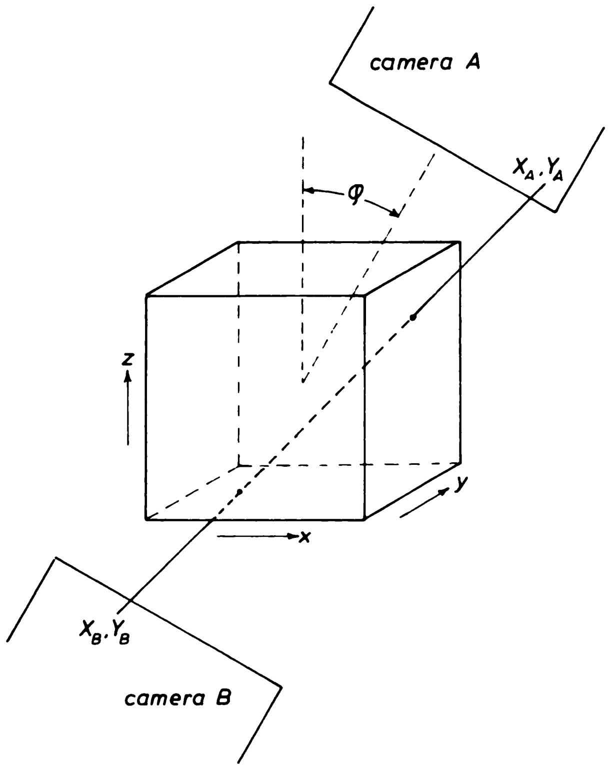

FIGURE 1
Coordinate system used in back projection process

signal is sent back to the electronics of the individual cameras in order to reset their electronics. The four position signals are connected to an ADC system installed in the processor box of a gamma-11 (PDP-11/34) computer system†. This ADC system is an AR-11 module† which comprises a 16-fold multiplexed, 10 bits ADC system. The analog to digital conversion is started by an external start pulse supplied by the coincidence electronics. The coincidence electronics also have the option for a hardware reconstruction of one focal plane, parallel to the detectors, if used in stationary mode (*5*). The normal nuclear medicine procedures can be followed when this hardware reconstruction option is used. This possibility is very useful when smaller animals like rats and hamsters are studied with positron emitting compounds.

The data acquisition software is a flexible and interactive code, which takes advantage of the real time properties of the RT-11 operating system. Also, the control of the rotation of the gantry is performed by this program. The digitized position signals are written to disk in buffers together with the time elapsed since the start of the data acquisition and the rotation angle of the Rota system. The camera system rotates, under program control, in steps of 3, 6 or 10 degrees, selectable on the console of the Rota system. If short-lived radionuclides like carbon-11 and nitrogen-13 are employed, the amount of activity at the beginning of the study, because of decay, is different from that at the end. The program has an option which recalculates the data acquisition time per step, based on the physical half-life and the time elapsed since the start of the data acquisition in order to obtain equal effective data acquisition times for all angles.

Due to the use of area detectors like scintillation cameras the reconstruction problem is three-dimensional instead of two-dimensional problem, as it is in the case of a ring system. In the back projection process after a rotating data acquisition, one first has to convert the measured camera coordinates into space-fixed coordinates. After this conversion, one has to find how the straight line, connecting the measured positions, travels through the volume defined for reconstruction. The coordinate system used for the reconstruction is shown in Fig. 1. The conversion from camera coordinates to space coordinates is given by:

$$X_1 = X_A \cos\phi + D/2 \sin\phi \tag{1}$$

$$Y_1 = Y_A \tag{2}$$

$$Z_1 = -X_A \sin\phi + D/2 \cos\phi \tag{3}$$

$$X_2 = X_B \sin\phi - D/2 \sin\phi \tag{4}$$

$$Y_2 = Y_B \tag{5}$$

$$Z_2 = -X_B \sin\phi - D/2 \cos\phi, \tag{6}$$

with (X_A, Y_A) and (X_B, Y_B) being the camera coordinates of camera A and B, respectively, and (X_1, Y_1, Z_1) and (X_2, Y_2, Z_2) the corresponding coordinates in space. The rotation angle is defined by ϕ. The detector separation is denoted by D. After this conversion, the plane of entrance and the direction of the line between the two coordinates have to be found. With this knowledge all volume elements traversed are incremented.

The response of a double-headed positron camera system is not uniform due to the change in solid angle with a change in source position (*3, 4, 5*). In a stationary system a correction for this nonuniform response can be applied by measuring the system response with a uniform source in the plane of interest. With a rotating camera system, a uniform response is required on mathematical grounds if the images obtained by back-projection have to be corrected for the system response by a deconvolution technique in Fourier space (*6*). To obtain a uniform response, a limited angle reconstruction option is built into the back projection program (*7-9*). The value of the new acceptance angle, smaller than the maximum acceptance angle, to be given to the program depends on the size of the object and the detector separation. By inspecting all events, whether they fall within this new acceptance angle or not, it is possible to obtain a uniform response for a certain part of the field of view of the camera system at the cost of some loss in sensitivity (*4, 10*). After the three-dimensional Fourier transform of the back projected images, the system response is divided out. The analytic expressions used for the system response of a double-headed positron camera were derived by Colsher (*11*). After the inverse Fourier transform the images are stored on disk as a standard patient study for further analysis with the nuclear medicine package.

The implementation of the back projection process and the three-dimensional Fourier deconvolution requires some consideration, since on a 16-bit computer, at maximum 2^{16} bytes = 64 kbytes can be addressed directly. In the back projection process it is necessary to process each event only once for reason of computing time. For this reason all volume elements, $64 \times 64 \times 16$ elements = 64 kwords = 128 kbytes, should be in the memory of the PDP-11/34. This is possible by using a

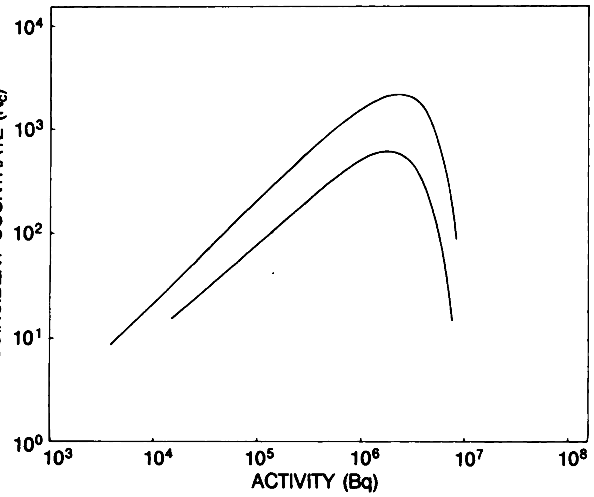

FIGURE 2
Coincident count rate as function of activity for small volume, upper curve, and large volume, lower curve

memory management unit by which the maximum addressable memory is increased to 256 kbytes. The lowest 64 kbytes are addressed directly and the remaining 192 kbytes have to be addressed indirectly by way of the memory management unit.

The naive estimation of the number of operations required to perform a discrete three-dimensional Fourier transform of $64\times64\times16$ data points is $(64\times64\times16)^2 = 4.3\times10^9$; one operation being a complex multiplication and one complex addition. The technique of the fast Fourier transform (FFT), exploiting the equal spacing of the samples and the periodicity of the integrand, reduces the number of operations and therefore the running time by orders of magnitude (*12*). Since the FFT has to be carried out in real numbers, to avoid large round-off errors, the amount of memory required just for the data array to perform the three-dimensional FFT is at least 264 kbytes, which exceeds the capacity of a PDP-11/34. For this reason the transformation is carried out in steps. First the two-dimensional FFT's of the 16 transverse section 64×64 matrices are performed and the results are stored on disk. For the transform in the third dimension the array consisting of a row of each of the 16 images is built up from the results on disk. After the transform the results on disk are updated and the next transform is performed. Since this involves the reading and writing of 50 Mbytes in total, special attention has been paid to the speed of the reading and writing. Since the time-consuming part of the three-dimensional transform is due to the limited amount of addressable memories, a computer with a virtual memory system would decrease the time required for the transform with a large factor. An array processor, coupled to a PDP-11/34, would decrease the execution time at most with a factor 2 since the time required for the input and output will remain the same.

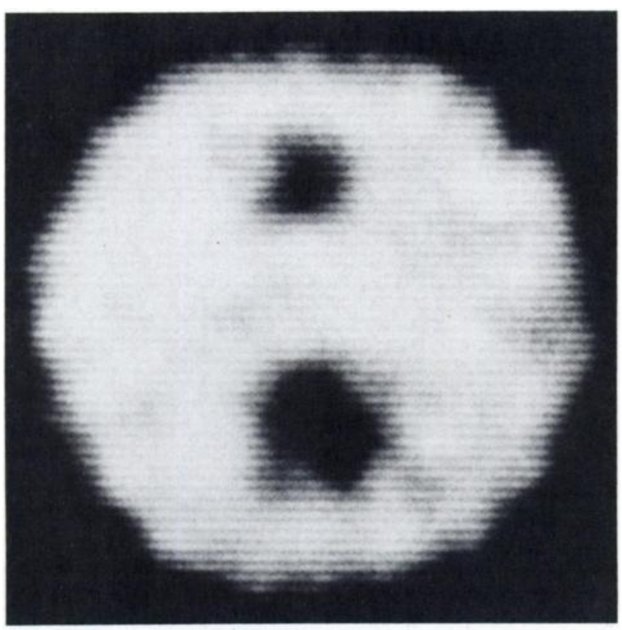

FIGURE 3
Transverse section image of cold spot phantom. Diameter of cold spots are 2 and 3 cm

After the transform, the system response is divided out. To avoid over-emphasizing of the noise and oscillations a window function with a smooth cutoff is used. Based on the results of Colsher and Townsend, a Hanning window function is applied (*9, 11*). After the inverse transform the images are available for inspection with help of the nuclear medicine package. The numerical stability of the deconvolution has been tested and showed differences in the order of 10^{-6} or less which are due to the limited presentation of real numbers in a computer system.

DISCUSSION

Spatial resolution

A number of factors affect the spatial resolution of the rotating positron camera. These factors are: (a) the intrinsic resolution of the individual cameras; (b) the finite range of positrons in matter; (c) the two annihilation quanta will not be exactly colinear; (d) a detector does not discriminate for the depth in the crystal at which the interaction takes place; (e) camera distortions; (f) mechanical stability of the rotating gantry; and (g) the calibration and off-sets of the position signals of the cameras.

The first five factors are identical for a stationary camera system, and their influence on the spatial resolution is described elsewhere (*3, 5*). Since in the ZLC cameras 3/8-in.-thick crystals are employed, the effect of the depth of interaction in the crystal is less pronounced as it was in a LFOV-based system. Moreover, due to the use of ZLC cameras, the camera distortions will be of much less importance. When the system is rotated, such distortions are especially likely to amplify themselves. By the use of ZLC based cameras this effect is minimized. According to the mechanical specifications of the system, the maxmum deviation from an inscribed circle is within 1 mm while the rotational position accuracy amounts to 0.5°. The maximum error due to a deviation of 1 mm from the inscribed circle is 1 mm in position determination. The possible error in the rotational accuracy, at a detector separation of 65 cm, can lead to errors of 3 mm in the position determination. The calibration of the cameras is an important factor since a 10 mV difference in position

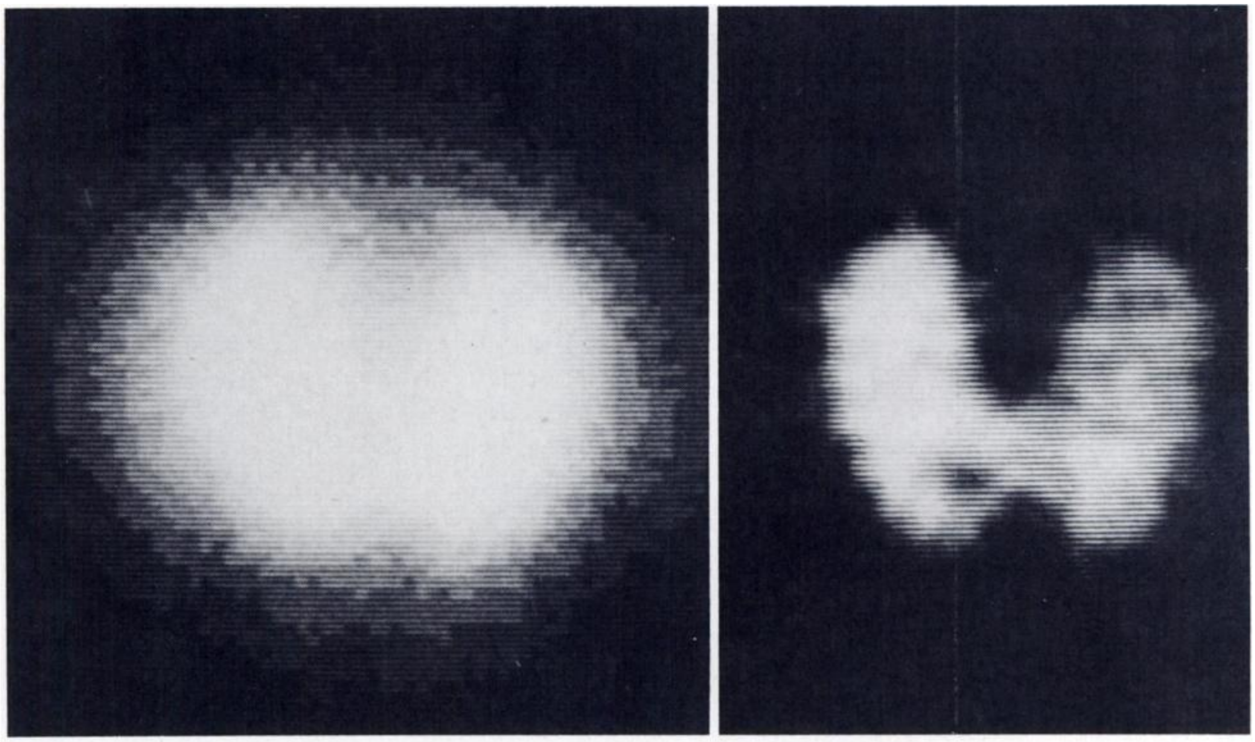

FIGURE 4
Transverse section images of thyroid phantom. On left-hand side image after back projection and right-hand side after deconvolution

signal amplitude leads to a 1 mm error in the position determination by the camera electronics. The spatial resolution measured with sodium-22 point sources amounted to 4.8 mm FWHM for the system in a stationary position. In the rotating system a spatial resolution of 7.5 mm FWHM was obtained in the transverse section image.

Time resolution

The energy signals of both cameras have been utilized, after pulse height discrimination by timing single channel analysers, to start and stop a time to pulse height converter. Multichannel analysis of the spectrum shows that the time resolution of the system varies between 7 and 12 ns FWHM depending on the amount of radioactivity between the cameras. Based on these values the resolving time window of the concidence electronics is normally set at 25 ns.

Sensitivity and count rate performance

The sensitivity of the system for a point source in the geometrical center is 100 cps/μCi (2.7 cps/kBq), at a detector separation of 50 cm. This value is in agreement with the calculated sensitivity. This calculation is based on the solid angles subtended and the photopeak efficiency of 3/8-in.-thick crystals for annihilation radiation.

The maximum coincident count rate obtainable is limited by the singles count rates in the detectors, due to the use of area detectors. The coincident count rate as function of the amount of activity has been measured for two extreme source geometries. The upper curve in Fig. 2 was obtained using a gallium-68 (^{68}Ga) solution with a volume of 10 cm^3 while for the measurement of the lower curve a uniform ^{68}Ga solution with a volume of 4,700 cm^3 was used. Both curves were obtained at a detector separation of 50 cm, and the energy window used ranged from 450 to 560 keV. Since the singles count rate is the limiting factor, the use of graded absorbers, as suggested by Muehllehner (*3*), will not only shift the maximum in the curves to the right but also show an increase in the maximum obtainable coincident count rate from a large volume (lower curve). Using graded absorbers consisting of 2 mm of lead followed by 1 mm of cadmium, to eliminate the x-rays generated in the lead, a maximum coincident count rate of 750 cps was reached with 5 MBq of ^{68}Ga uniformly distributed in a volume of 4,700 cm^3.

The measured coincident count rate consists of three components: the "true" event rate, the "scattered" event rate and the "accidental" event rate. True events are unscattered annihilation pairs detected within the coincidence time window and within the energy window. Scattered events are annihilation pairs in which one or both have been scattered before detection within the time and energy windows. Accidental events are two gamma-rays originating from different positron-electron pairs but detected within the time and energy windows. Since useful spatial information is only represented by the true events, insight into the extent of the scattered and accidental events is essential. To measure the number of scattered

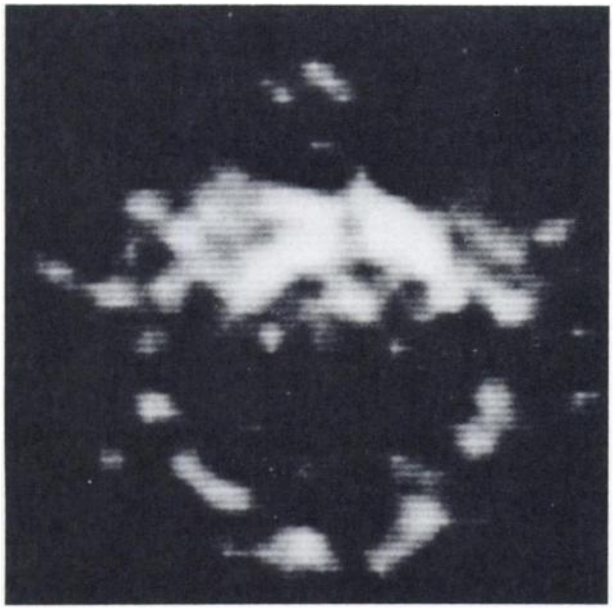

FIGURE 5
Brain image of dopamine receptors with ^{11}C-labeled methylspiperone in normal volunteer

events a cold spot phantom was used (*13*). By using a relatively low amount of radioactive material the number of accidental events can be neglected since their number is proportional to the singles count rates and the time window used. The cold spot phantom used was a 18-cm-long cylinder, 12 cm in diam, with two cold spots, parallel to cylinder axis, center distance 5.5 cm, with diameters of 2 and 3 cm, respectively. Data acquisition was performed in steps of 10° and the data acquisition time per step was recalculated on the basis of the half-life of ^{68}Ga (68.3 min). In total 500,000 events were recorded. A transverse section image of the phantom is shown in Fig. 3. The amount of scattered events was determined to be 15% by fitting the background with polynomials. This percentage is of the same magnitude as reported for ring systems (*13, 14*), although comparison may be difficult due to differences in detector and collimator geometries. The measurements of Derenzo et al. (*14*) also show that the scatter fraction is not strongly dependent on the energy window used. If this observation is also true for scintillation camera-based positron imaging devices, the true coincident count rate from extended sources can be raised by approximately a factor of 5 without increasing the singles count rate (*15*).

Uniformity of the system

As already mentioned, a double-headed positron camera has a rather nonuniform response. A limited angle reconstruction technique in the back projection process is able to remove this nonuniformity (*9*). Experimentally, the uniformity of the system as function of the acceptance angle was verified using a 50 cm long, 1.7-cm-diam line source, filled with ^{68}Ga ($t_{1/2}$ = 68.3 min), at different positions between the cameras. The camera separation was 50 cm and the volume used for reconstruction of the images was 40 × 40 × 40 cm^3, divided into 16 transverse sections. From these measurements it was concluded that a uniform response over a distance of ~ 20 cm along the rotation axis can be obtained at the cost of some loss in sensitivity. The actual value of this loss is dependent on object size and detector separation.

IMAGING EXAMPLES

To show the potential of the system, a few images of phantom and patient studies are shown. As a second phantom study, the transverse section image of a thyroid phantom is shown in Fig. 4. On the left hand side, the back projected image is shown, and on the right hand side, the image after the subsequent deconvolution. The volume defined for reconstruction was 15×15×15 cm^3. The cold and hot spots of 11 mm diam are clearly visible. The 8-mm-diam cold spot in the upper part of the right lobe is not visible or barely visible. In Fig. 5, an image of the brain, showing the dopamine receptors in the striata of a normal volunteer is shown. The image was obtained 60 min after injection of 5 mCi (185 MBq) ^{11}C-labeled 3-N-methylspiperone. The data acquisition was performed in steps of 10°. The data acquisition time was 1 min per step and was adjusted per step for the physical decay of ^{11}C, ($t_{1/2}$ = 20.3 min). The total time required for data acquisition was 35 min. The same data acquisition procedure was used for obtaining the brain image shown in Fig. 6. Data acquisition was started 10 min after injection of 2.5 mCi (90 MBq) D,L-(1-^{11}C)-tyrosine. The accumulation of the amino acid in the grade III glioma in the lower right part of the brain is obvious.

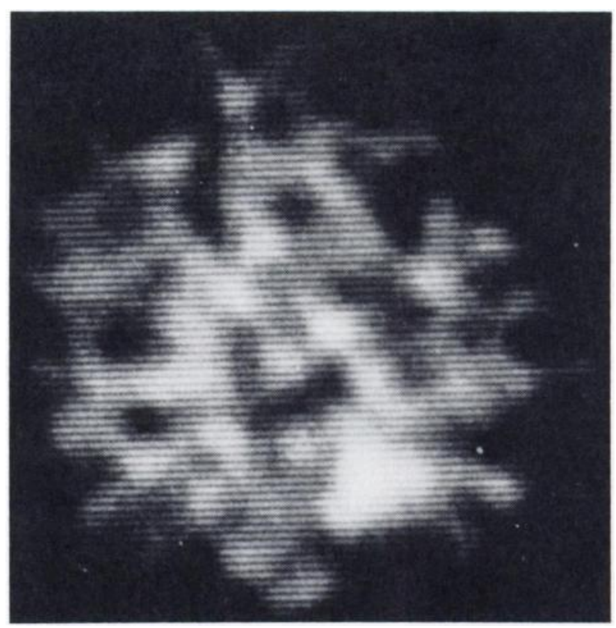

FIGURE 6
Brain image with D,L-(1-^{11}C)-tyrosine. Amino acid accumulates in grade III glioma in right lower part of brain

CONCLUSIONS

It was shown that a rotating double-headed camera system, operating in a coincidence mode, is able to deliver high quality transverse section images of positron emitting radiopharmaceuticals in both phantom and patient studies.

It was also shown that the rather complex three-dimensional back projection and deconvolution processes can be carried out on a small general purpose computer system. The time required for the back projection process is rather long but by using the batch handler, available under the RT-11 operating system, this can be conducted overnight.

Due to the use of area detectors there are limitations on the singles count rate the system can handle. Some flexibility is obtained when graded absorbers, which decrease the singles count rate due to the low-energy (scattered) gamma rays to a much larger extent than the singles count rate due the 511 keV photo-peak events, are available. The limitation set by the time required to perform a rotating data acquisition implicates that all studies are limited to steady-state processes.

An advantage is of course that a single photon imaging device can be converted into a positron imaging system. In this way a new option is added to an already present system, while switching back to single photon imaging means only the remounting of the collimators and the switching off of the coincidence electronics.

FOOTNOTES

*Siemens Gammasonics B.V., The Netherlands.
†Digital Equipment Corporation, USA., Maynard, MA.

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