

EUROPEAN ORGANIZATION FOR NUCLEAR RESEARCH

CERN/DATA HANDLING DIVISION
DD/82/15
15 October 1982

IMAGE RECONSTRUCTION FOR A ROTATING POSITRON TOMOGRAPH

D. Townsend¹, B. Schorr², A. Jeavons³, R. Clack¹,
R. Magnanini¹, P. Frey¹, A. Donath¹ and A. Froidevaux³

Presented at the
IEEE Nuclear Science Symposium
Washington DC, 20-22 October 1982

-
1. Hôpital Cantonal, Geneva, Switzerland.
 2. DD Division, CERN, Geneva, Switzerland.
 3. EP Division, CERN, Geneva, Switzerland.

Abstract

A high-resolution positron tomograph consisting of two high-density avalanche chambers mounted on a rotating gantry has been installed in the Nuclear Medicine Department of the Cantonal Hospital in Geneva. Positron annihilation data are collected from six detector positions and back-projected in real time to form a single three-dimensional image. Up to 16 transverse or longitudinal sections may be imaged simultaneously. The derivation of the appropriate deconvolution filter is described in this paper, and the first images taken in the transverse mode are presented. It is shown that quantitative information can be obtained from such images.

1. Introduction

The requirement of adequate angular sampling in order to obtain accurate quantification of positron emission tomography (PET) images is well known. The effects of limited or incomplete angular information have been discussed by several authors^{1,2}, and have been shown to lead to artefacts, distortions and loss of quantification in the PET image. Techniques³ to estimate the missing angular information from the measured data have not proved very successful in practice¹, due mainly to the noisy nature of PET images.

In particular, the limited angle reconstruction problems associated with the large area, dual detector, stationary positron camera have been extensively studied. It has been shown⁴ that the appropriate deconvolution filter has a cone or pyramid of zeros in frequency space resulting from frequencies not measured by the camera. As a consequence, the reconstructed image is considerably distorted, particularly along the direction perpendicular to the detectors.

To overcome these problems, a dual detector system has been mounted on a rotating support that allows full angular sampling through 180° . It is therefore possible to measure projections not previously available to the stationary system, projections corresponding to the missing frequencies. To take advantage of the rotational capability, modifications to the stationary system data acquisition and image reconstruction software were necessary.

This paper contains a description of the rotational image reconstruction software, and in particular the back-projection algorithm and the derivation of the three-dimensional deconvolution filter. Some simple phantom measurements made with the new system illustrate the image reconstruction process and indicate that high-resolution, quantitative imaging should be feasible.

2. The Rotating Camera

A prototype positron camera based on a pair of $20\text{ cm} \times 20\text{ cm}$ square, high-density avalanche chambers (HIDAC) has been mounted on a circular rotating gantry of diameter 1.2 m and installed in the Department of Nuclear Medicine of the Cantonal Hospital in Geneva. The chamber separation is variable, and is currently 34 cm. In rotational mode, six views are obtained at

30° increments to give the full 180° . At each position, annihilation data are accepted within a cone of $\pm 15^\circ$, such that the camera has a uniform point response throughout a sphere of diameter 11 cm, positioned centrally. Data rejection as a result of the cone constraint varies from 70% at the centre of the sphere to 50% at the edges. On the average, about 60% of the data are rejected as falling outside the cone in a typical data run.

The camera sensitivity for a point source in air is approximately 10 cps/ μCi , and, with a time resolution of 20 ns, count rates of 4 kHz are achieved with an accidental coincidence rate of 25%. Thus, to obtain about one million events in a three-dimensional image, 400,000 counts are collected at each angular position which, at an average rate of 3 kHz, corresponds to an imaging time of 2.2 min. The total time is therefore 13 min for one million counts in the image, of which about 250,000 are accidentals, distributed uniformly throughout the imaging volume.

In the case of positron activity distributed more or less uniformly throughout the 11 cm diameter sphere, a data collection time of 13 minutes will result in a 16% statistical error in voxels of dimension $2\text{ mm} \times 2\text{ mm} \times 8\text{ mm}$. The background from accidentals will be about 5%, since they are distributed throughout the reconstruction matrix. In practice, of course, scatters will make this situation somewhat worse by reducing the good event rate and increasing the background. A discussion of the camera response to scattered radiation may be found elsewhere⁵.

In a previous publication⁶, details of the microprocessor-based data acquisition system were covered, and they will not be repeated here. Instead, in the following section, details of the modifications for the back-projection algorithm are given, as well as the derivation of the appropriate three-dimensional deconvolution filter required in the reconstruction of rotational images.

3. Rotational Image Reconstruction

The back-projection procedure in the case of the dual-detector, stationary system consisted of intersecting annihilation event lines with a set of planes parallel to the detectors. This situation is shown in fig. 1 as detector position 2, where the x-axis is

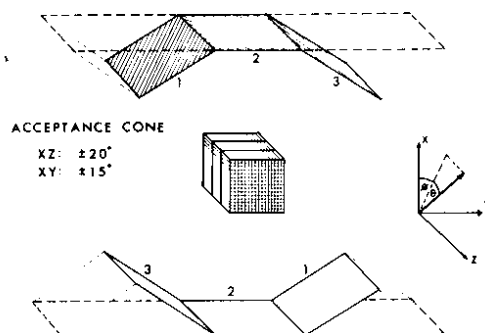


Fig. 1 Coordinate system for the rotational camera.

perpendicular to the detectors and the reconstruction planes are parallel to the yz -plane. If the detectors are now rotated about the z -axis to the positions marked 1 and 3, the back-projection algorithm must accept the additional data.

In order to preserve the plane intersection approach, and to avoid the summing of reconstruction matrices taken at different angles, the back-projection planes are kept fixed as the detectors are moved to positions 1 and 3. Therefore, events detected in these positions still intersect planes parallel to the yz -plane, so that the basic algorithm remains unchanged. Instead, the coordinates of events detected in positions 1 and 3 are transformed into the base system as defined by detector position 2. This procedure is equivalent to extending the stationary detectors in the y -direction, as illustrated in fig. 1.

3.1 The back-projection transformation

The details of this transformation are summarized in fig. 2. Detectors in a base position (a) are rotated about a point x_r, y_r through angle θ_r to position b, as shown in fig. 2a. An annihilation event occurring at O is detected by the rotated camera. The intersection points of this event line with the detectors in position a extended as shown are found by first expressing the detection points in terms of the base coordinate system, i.e. the points (x_1, y_1, z_1) and (x_2, y_2, z_2) , and then using the direction of the event to extrapolate from these points to y'_1, z'_1 and y'_2, z'_2 on the detectors in the base position. The transformation in the plane of the rotation (xy) is shown in fig. 2b, and the transformation in the plane perpendicular to the rotation (xz) is shown in fig. 2c. The coordinates

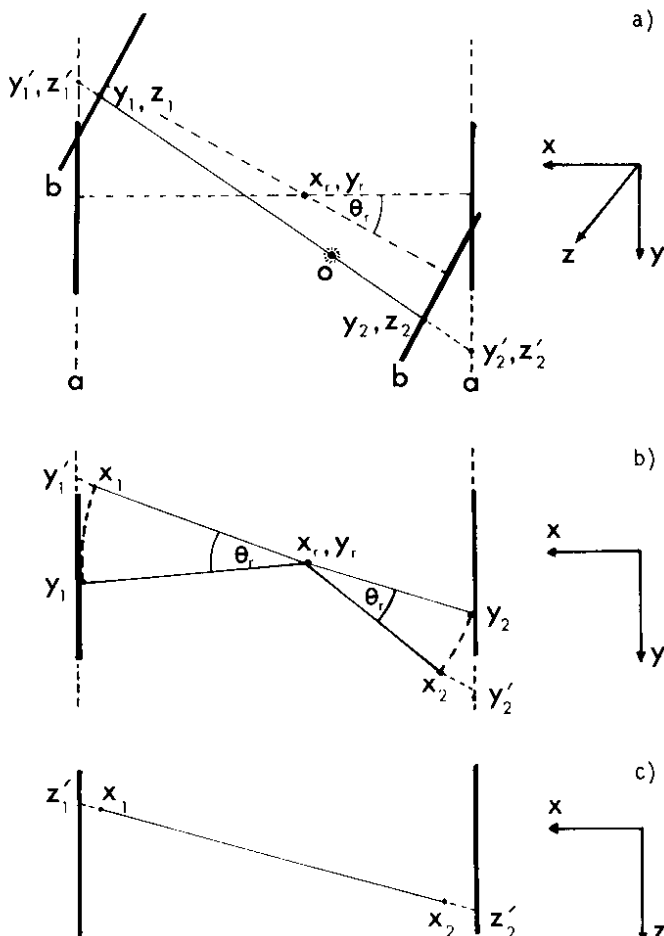


Fig. 2 Coordinate transformation for back-projection.

x_1 and x_2 are found by a rotation of θ_r about a point (x_r, y_r) for all possible detector coordinates y_1 and y_2 , as shown in fig. 2b. The x -coordinate may therefore be tabulated as a function of y_1, x_r, y_r , and θ_r . Thus, once x_1 and x_2 are known, z'_1 and z'_2 are found immediately by extrapolation because z_1 and z_2 , lying in the plane perpendicular to the rotation, are the same as the detected z -coordinates. The detected y -coordinates must first be transformed to y_1 and y_2 . In practice, this is done by first translating the event line to the centre of rotation, rotating through θ_r , and then applying a reverse translation. The final step is then to extrapolate from x_1, y_1 and x_2, y_2 to give y'_1 and y'_2 on the extended detectors. Apart from table look-up, these transformations add a further three multiplications and two divisions to each event. The transformed coordinates are then fed directly into the original back-projection algorithm.

The centre of rotation x_r, y_r lies close to the mid-point of the system, but for mechanical reasons it is not possible to arrange for the two to coincide. Instead, it is necessary to measure the centre of rotation by mapping a sequence of point sources until the exact position that remains invariant is found. This may be done, in practice, to an accuracy of 1 mm in x_r and y_r . To the same accuracy, it was found that the axis of rotation was parallel to the detectors, i.e. the detectors are perpendicular to the plane of rotation.

To complete the full rotation, the detectors are moved to positions 4, 5, and 6 (see fig. 3) at angles of $+60^\circ$, $+90^\circ$, and $+120^\circ$, respectively, to the x -axis. At these positions, a similar back-projection procedure is adopted, with the exception that the intersection planes are parallel to the xz -plane, i.e. parallel to the detectors in position 5 and rotated by 90° from the back-projection in positions 1, 2, and 3. By ensuring that the pixels on xy -planes are square and that the centre of the three-dimensional reconstruction matrix lies at the centre of rotation, and since back-projection is a linear process, the two independent back-projections for positions 1, 2, 3 and 4, 5, 6 may be summed to form a single back-projected image corresponding to full rotation.

Finally, by knowing the point response function of the procedure described above, Fourier space deconvolution yields the reconstructed image. Computation of the deconvolution filter from the point response function is not always easy, and therefore a general method

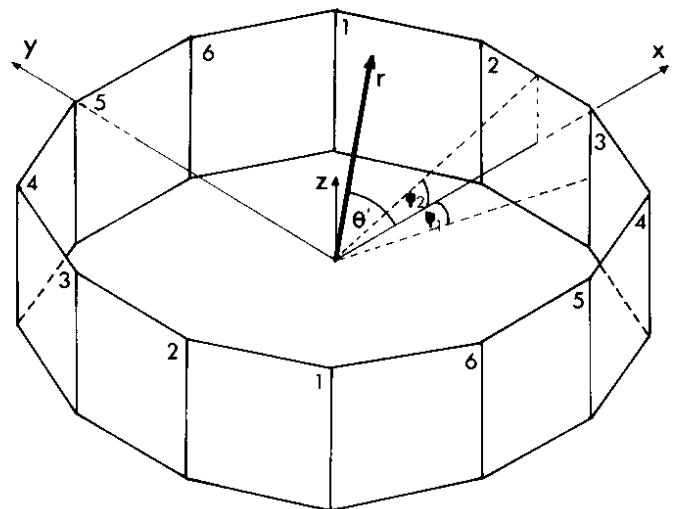


Fig. 3 Six detector positions for full rotation: ψ_1 = acceptance angle in xy -plane; ψ_2 = acceptance angle in xz -plane.

has been evolved to simplify this computation. The full details of this method may be found elsewhere⁷, but the essential features are summarized in the next section. Application of this method to compute the filter for the rotational back-projection process described above is outlined in section 3.3.

3.2 General filter computation

Let $\hat{a}(x,y,z)$ be the back-projected image of an unknown activity distribution, $a(x,y,z)$. The point response function h of the imaging system is assumed to be stationary, such that a and $\hat{a}$ are related by

$$\hat{a} = h * a, \quad (3.1)$$

where $*$ denotes three-dimensional convolution. Deconvolution of equation (3.1) may be obtained by applying the three-dimensional Fourier transform, yielding

$$\hat{A} = H \cdot A, \quad (3.2)$$

where $\hat{A}$, H , and A are the corresponding Fourier transforms. Assuming H is known and A is obtained from a discrete Fourier transform of the data $\hat{a}$, the activity distribution a is computed from the inverse Fourier transform of $\hat{A}/H$. As explained previously, with incomplete angular sampling, not all frequencies are measured, and it is found in such a case that H vanishes identically in some three-dimensional region in frequency space. Under these circumstances, only an approximation to a can be found.

To apply this reconstruction method, it is necessary to know H , the three-dimensional Fourier transform of the point response function. Since different imaging systems have different response functions, a method has been devised that allows the computation of a fairly general class of filter functions. Suppose the point response function in polar coordinates may be expressed in the form

$$h(r, \theta, \phi) = \frac{d(\theta, \phi)}{r^2}, \quad \text{for } -\infty < r < \infty, (\theta, \phi) \in \Omega, \quad (3.3)$$

where Ω is a closed subset of

$$E_{\theta, \phi} = \{(\theta, \phi); -\pi/2 \leq \theta \leq \pi/2, -\pi/2 \leq \phi \leq \pi/2\}$$

and where polar and Cartesian coordinates are related by

$$\begin{aligned} x &= r \cos \phi \cos \theta \\ y &= r \sin \phi \cos \theta \\ z &= r \sin \theta. \end{aligned} \quad (3.4)$$

The function $d(\theta, \phi)$ in eq. (3.3) is termed the detector function because it allows for the detector acceptance and angular factors occurring in the back-projection. It is assumed that

$$d(\theta, \phi) = d(-\theta, \phi) = d(\theta, -\phi) \quad (3.5)$$

which is sufficiently general for all practical purposes.

Using polar coordinates in frequency space, defined similarly to eqs. (3.4), the Fourier transform of the point response function is given by

$$H(R, \Theta, \Phi) = \iint_{\Omega} d(\theta, \phi) \cos \theta \, d\theta \, d\phi \int_{-\infty}^{\infty} e^{-2\pi i r R (\cos(\Phi - \phi) \cos \theta \cos \Theta + \sin \theta \sin \Theta)} dr$$

$$\text{for } -\infty < R < \infty, -\pi/2 \leq \Theta \leq \pi/2, -\pi/2 \leq \Phi \leq \pi/2, \quad (3.6)$$

where the integration in (3.6) is to be considered within the theory of generalized functions⁸.

It is shown elsewhere⁷ that, under the condition eq. (3.5) and for $\Omega \in \Omega_0$, with

$$\Omega_0 = \{(\theta, \phi); -\theta_0 \leq \theta \leq \theta_0, -\phi_0 \leq \phi \leq \phi_0\}, \quad 0 < \theta_0 \leq \pi/2, 0 < \phi_0 \leq \pi/2,$$

the filter function H is given by

$$H(R, \Theta, \Phi) = \frac{D(\Theta, \Phi)}{|R|} \quad (3.7)$$

where D satisfies

$$D(-\Theta, \Phi) = D(\Theta, -\Phi) = D(\Theta, \Phi) \quad (3.8)$$

and is given by

$$D(\Theta, \Phi) = \int_{\Phi - \phi_0}^{\Phi + \phi_0} s(\phi; \theta_0, \Theta) g(\phi; \Theta, \Phi) \, d\phi, \quad (3.9)$$

$$\text{for } 0 < \Theta < \pi/2, 0 \leq \Phi \leq \pi/2.$$

$$D(\Theta, \Phi) = \begin{cases} 0 & 0 \leq \Phi < \pi/2 - \phi_0 \\ 2 \int_0^{\theta_0} d(\theta, \pi/2 - \Phi) \, d\theta & \pi/2 - \phi_0 \leq \Phi \leq \pi/2 \end{cases} \quad (3.10)$$

$$D\left(\frac{\pi}{2}, \Phi\right) = 2 \int_0^{\phi_0} d(0, \phi) \, d\phi \quad 0 \leq \Phi \leq \pi/2, \quad (3.11)$$

with

$$s(\phi; \theta_0, \Theta) = \frac{1}{2} \{ \text{sign}(\cos \phi + \tan \theta_0 \tan \Theta) - \text{sign}(\cos \phi - \tan \theta_0 \tan \Theta) \} \quad (3.12)$$

$$g(\phi; \Theta, \Phi) = \frac{\tan \Theta}{\cos \Theta} \frac{d(\tan^{-1}(\cos \phi / \tan \Theta), \Phi - \phi)}{\tan^2 \Theta + \cos^2 \phi}. \quad (3.13)$$

In ref. 7, it is also shown that

$$0 < \Theta < \pi/2 - \theta_0,$$

$$D(\Theta, \Phi) = 0,$$

$$0 \leq \Phi \leq \cos^{-1}(\tan \theta_0 \tan \Theta) - \phi_0,$$

whilst for other Θ, Φ values the function $D(\Theta, \Phi)$ is positive if the full angular ranges $0 \leq \theta \leq \theta_0$ and $0 \leq \phi \leq \phi_0$ are included in Ω , although θ and ϕ are not necessarily independent. It then follows that $D(\Theta, \Phi)$ is never zero in a two-dimensional region if either $\phi_0 = \pi/2$ or $\theta_0 = \pi/2$, and in either case an exact reconstruction is possible. If, however, both of $\phi_0 < \pi/2$ and $\theta_0 < \pi/2$, only an approximate, limited-angle reconstruction of a may be obtained.

3.3 The filter for the rotational system

The method outlined in the previous section may be used to compute the appropriate deconvolution filter corresponding to the rotational back-projection procedure described in section 3.1. It is known⁹ that for the stationary, dual-detector system, back-projection

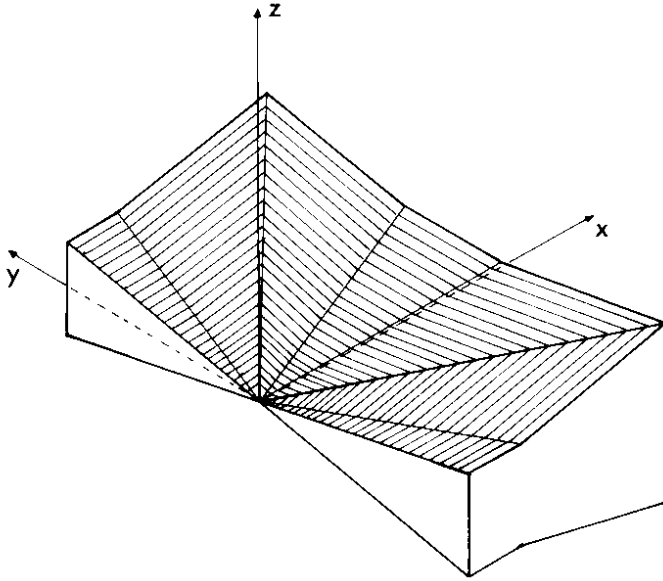


Fig. 4 Filter integration volume.

by plane intersections yields a point response function of the form $\cos \theta'/r^2$, where θ' is the angle between the event line and the x-axis. It is clear that the back-projection procedure described in section 3.1 will yield a point response function also of the form $\cos \theta'/r^2$, where, for positions 1, 2, and 3, θ' is the angle between the event line and the x-axis, whilst for positions 4, 5, and 6, θ' is the angle between the event line and the y-axis (see fig. 3). The support of the response function [i.e. the region in which $h(r, \theta, \phi)$ does not vanish] is illustrated in fig. 4.

It can be shown by a straightforward but lengthy derivation, that the detector function for this rotational system may be written:

$$\begin{aligned} d(\theta, \phi) &= \cos \theta \cos (\phi + \beta_i), \\ -\alpha_i &\leq \phi \leq \alpha_{i+1}, \\ -\theta_i(\phi) &\leq \theta \leq \theta_i(\phi), \quad i = 1, 2, \dots, 7, \end{aligned} \quad (3.14)$$

where

$$\begin{aligned} \theta_i(\phi) &= \tan^{-1} \left\{ \tan \psi_2 \cos (\phi + \beta_i) \right. \\ &\quad \left. + \gamma_i \left[\cos (\phi + \delta_i) - \tan \psi_1 (\cos (\phi + \beta_i)) \right] \right\}, \\ i &= 1, 2, \dots, 7. \end{aligned} \quad (3.15)$$

The angles ψ_1, ψ_2 are as shown in fig. 3, and

$$\alpha_1 = -\pi/2, \quad \alpha_2 = -5\pi/12, \quad \alpha_3 = -\pi/4, \quad (3.16)$$

$$\alpha_4 = 0, \quad \alpha_{4+j} = -\alpha_{5-j}, \quad j = 1, 2, 3, 4$$

$$\beta_1 = \beta_2 = \pi/2, \quad \beta_3 = \beta_4 = \beta_5 = 0, \quad \beta_6 = \beta_7 = -\pi/2$$

$$\gamma_1 = 0, \quad \gamma_2 = \gamma_3 = 1/2, \quad \gamma_4 = 0, \quad \gamma_5 = \gamma_6 = 1/2, \quad \gamma_7 = 0$$

$$\delta_1 = \delta_2 = 0, \quad \delta_3 = \pi/2, \quad \delta_4 = \delta_5 = 0, \quad \delta_6 = -\pi/2, \quad \delta_7 = 0. \quad (3.17)$$

Calculating the integration in eq. (3.9) for the detector function (3.14), it is not difficult to show that $D(\theta, \phi)$ may be written:

$$D(\theta, \phi) = \sum_{i=1}^7 \int_{\alpha_i}^{\alpha_{i+1}} s(\phi - \phi_i(\phi), \theta) g_i(\phi - \phi_i(\phi), \theta) d\phi, \quad (3.18)$$

where

$$g_i(\phi; \theta, \phi) = \frac{\tan \theta \cos (\tan^{-1} (\cos \phi / \tan \theta)) \cos (\phi + \beta_i - \phi)}{\cos \theta \tan^2 \theta + \cos^2 \phi} \quad (3.19)$$

for $i = 1, 2, \dots, 7$, $0 < \theta < \pi/2$, and $0 \leq \phi \leq \pi/2$.

Using a result from ref. 7, it can be shown that the indefinite integral of g_i in eq. (3.18) is given by

$$\begin{aligned} G_i(\phi; \theta, \phi) &= \int g_i(\phi; \theta, \phi) d\phi \\ &= \frac{\sin (\tan^{-1} (\cos \phi / \tan \theta))}{\cos \theta} \\ &\quad \times (\sin^2 \theta \cos (\phi + \beta_i) \tan \phi - \sin (\phi + \beta_i)). \end{aligned} \quad (3.20)$$

Finally, a detailed study of the functions $s(\phi - \phi_i(\phi), \theta)$ from eq. (3.18) which are defined in eqs. (3.12) and (3.15), shows that the integration limits in eq. (3.18) are given as follows. Let

$$c_i^{\pm} = -\tan^{-1} \frac{\cos \phi \pm (\epsilon_i \cos \beta_i + \gamma_i \cos \delta_i) \tan \psi_2 \tan \theta}{\sin \phi \mp (\epsilon_i \sin \beta_i + \gamma_i \sin \delta_i) \tan \psi_2 \tan \theta},$$

with

$$\epsilon_i = 1 - \gamma_i \tan \psi_1$$

for $i = 1, 2, \dots, 7$. The integration limits are then

$$\begin{aligned} \phi_i &= \max (\alpha_i, \min (c_i^-, c_i^+)), \\ \phi_{i+1} &= \min (\alpha_{i+1}, \max (c_i^-, c_i^+)), \end{aligned} \quad (3.21)$$

for $i = 1, 2, \dots, 7$. The final explicit expression for $D(\theta, \phi)$ may then be written, using eqs. (3.18), (3.20), and (3.21) and the fact that the function s takes the value 1 within the integration limits (3.21):

$$D(\theta, \phi) = \sum_{i=1}^7 (G_i(\phi - \phi_{i+1}; \theta, \phi) - G_i(\phi - \phi_i; \theta, \phi)) \quad (3.22)$$

for $0 < \theta < \pi/2$, $0 \leq \phi \leq \pi/2$. (The prime on the summation symbol means that there is no contribution to the sum for those i for which $\phi_i \geq \phi_{i+1}$.)

The limiting cases $\theta = 0$ and $\theta = \pi/2$ are easily obtained from eqs. (3.10), (3.11), (3.15), (3.16), and (3.17):

$$\begin{aligned} D(0, \phi) &= 2 \sin \theta_i (\pi/2 - \phi) \cos (\pi/2 - \phi + \beta_i), \\ \pi/2 - \alpha_{i+1} &\leq \phi \leq \pi/2 - \alpha_i, \end{aligned} \quad (3.23)$$

$$i = 4, 5, 6, 7,$$

$$D(\pi/2, \phi) = 4 \sin \pi/4 = 2\sqrt{2}, \quad 0 \leq \phi \leq \pi/2, \quad (3.24)$$

where θ_0 has to be replaced by $\theta_i(\pi/2 - \phi)$ and ϕ_0 by $\pi/2$ in the equations (3.10) and (3.11).

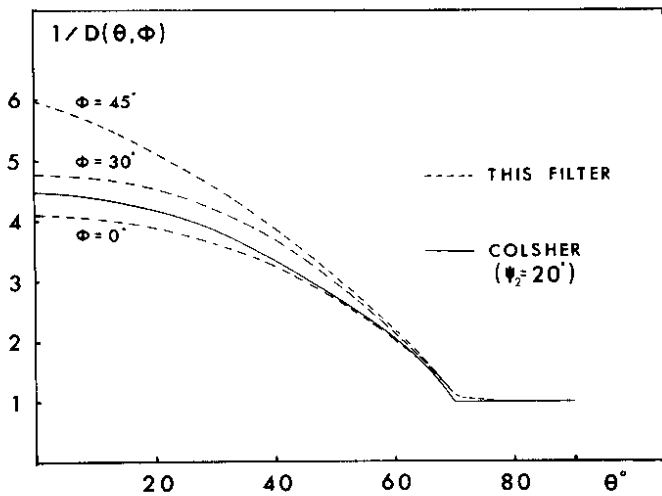


Fig. 5 Angular dependence of the rotational filter.

In fig. 5, $1/D(\theta, \phi)$ computed from eqs. (3.22), (3.23), and (3.24) is plotted as a function of θ for $\phi = 0^\circ$, 30° , and 45° , normalized to unity at $\theta = \pi/2$. Also plotted with the same normalization is a filter given by Colsher¹⁰ for a rotating positron camera. In his case, however, the back-projected image was formed by intersecting events with three-dimensional voxels as opposed to two-dimensional pixels on parallel planes, as considered here. The point response function in that case behaves as $1/r^2$ rather than $\cos \theta'/r^2$, and then the corresponding filter is independent of ϕ .

4. Transverse Imaging Results

Previous results from the stationary system have been presented as longitudinal images on planes parallel to the yz-plane (see fig. 1). Each such section is a 128×128 array of pixels, and up to 16 sections in the x-direction could be simultaneously imaged. It was shown¹, however, that because of the limited-angle artefacts, useful quantitative information could not be obtained from these images.

With the rotational system, transverse imaging becomes possible and the images presented here are on sections parallel to the xy-plane (fig. 1). Again, each image is a 128×128 array of pixels, but the 16 sections are taken consecutively along the z-axis, the axis of rotation. A future increase in memory capacity will make it possible to image 128 sections simultaneously. The section thickness in the stationary system depended strongly on the maximum angle of view (i.e. ψ_1 and ψ_2), while for the rotational system it is a trade-off between the spatial dimensions of the radioisotope activity, the photon statistics, and the available computer memory.

The first image (fig. 6) is that of a transverse section through a point source of activity close to the mid-point. In this and subsequent images, the x-direction is from top to bottom and the y-direction from left to right; the z-direction is perpendicular to the plane of the image. The vertical dimension of the isometric plot is proportional to the number of counts in the pixel. The back-projection is shown in fig. 6a and the reconstruction in fig. 6b for a pixel size of $1 \text{ mm} \times 1 \text{ mm}$ and an effective section thickness of 4 mm. After reconstruction, the 2 mm point source of ^{22}Na has a full width at half maximum of less than 4 mm.

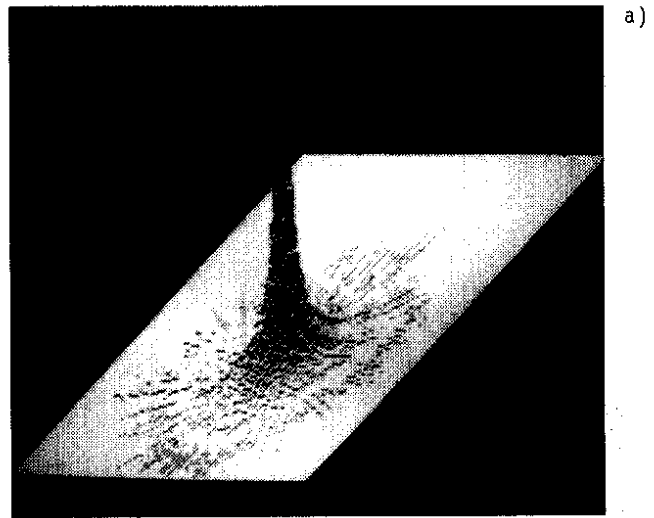
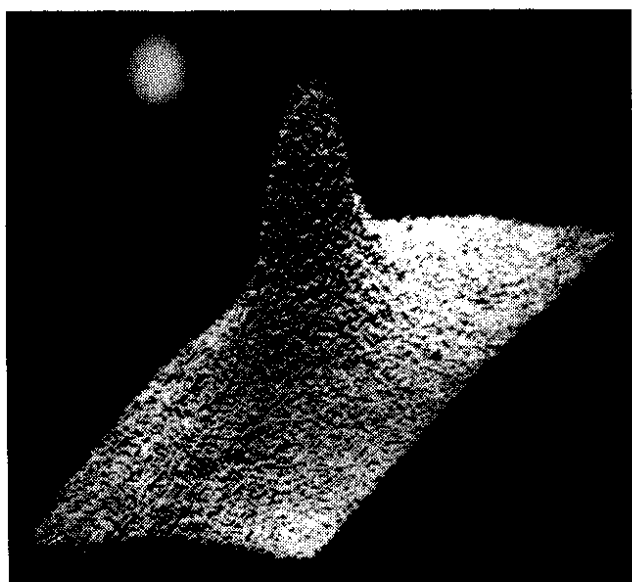


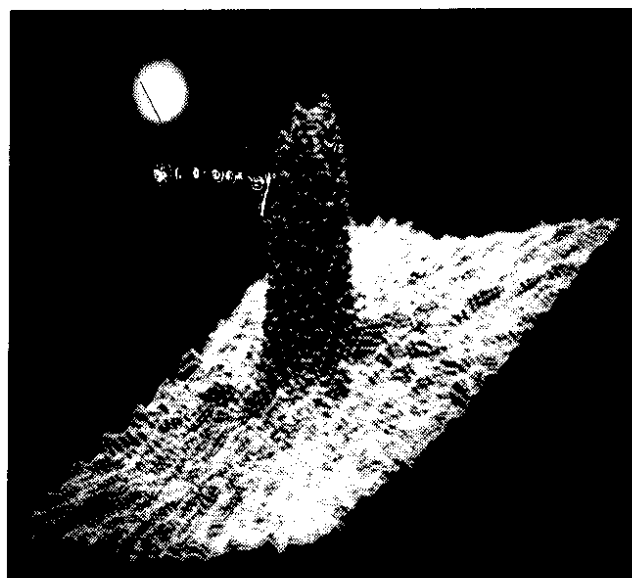
Fig. 6 Transverse section through a point source: a) back-projection; b) reconstruction.

The second example (fig. 7) shows an 8 mm section through a cylinder uniformly filled with ^{68}Ga activity. The cylinder has a diameter of 5 cm and is placed in the tomograph with the long axis parallel to the axis of rotation. The pixel size is $2 \text{ mm} \times 2 \text{ mm}$ and the image contains approximately 100,000 counts. The back-projection (tomogram) is shown in fig. 7a and the reconstruction in fig. 7b. The statistical noise seen in the tomogram is amplified by the filtering process and appears in fig. 7b surrounding the cylinder. In fig. 7c a profile taken through the tomogram (left) and reconstruction (right) clearly demonstrates the improvement in the image due to the filtering. The dip in the centre of the cylinder is a result of uncorrected photon attenuation.

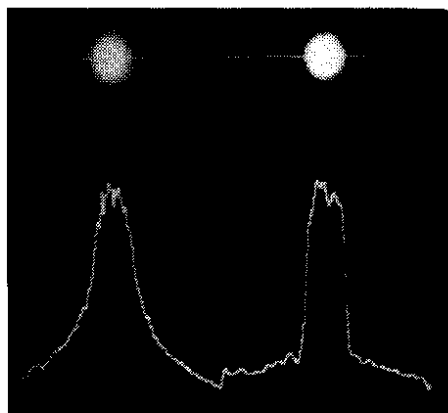
A more complex example is shown in fig. 8. The Picker thyroid phantom filled with 800 μCi of ^{68}Ga was placed centrally in the tomograph in a transverse plane perpendicular to the axis of rotation. The back-projection of a single 4 mm thick section containing about 200,000 counts is shown in fig. 8a with a pixel size of $2 \text{ mm} \times 2 \text{ mm}$; the corresponding reconstruction is shown in fig. 8b. The filtering greatly improves the image contrast and, although the 6 mm cold spot is



a)

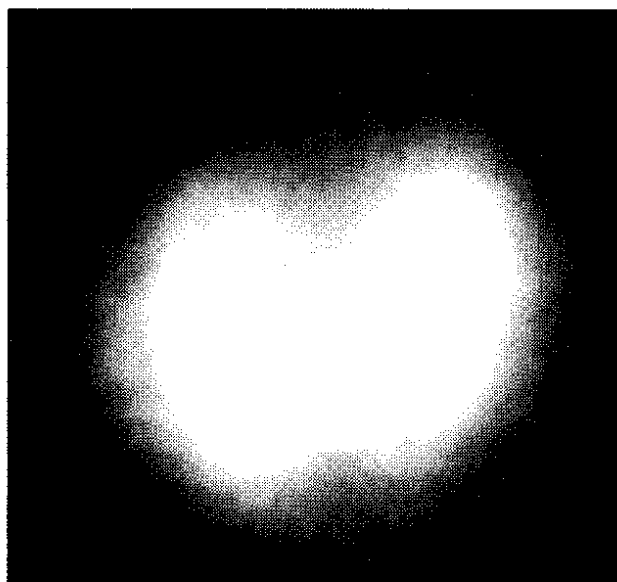


b)

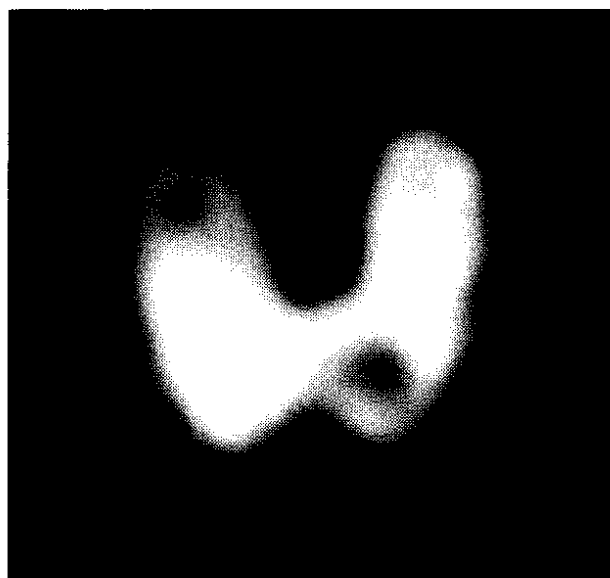


c)

Fig. 7 Transverse section through a cylinder uniformly filled with activity: a) back-projection; b) reconstruction; c) comparison of a profile through (a) and (b).



a)



b)

Fig. 8 Picker thyroid phantom; one section; ^{68}Ga activity: a) back-projection; b) reconstruction.

barely resolved, the larger 11 mm cold spots are clearly seen. The characteristic hot spot, usually seen towards the bottom of the left lobe, does not appear as such in this high-resolution transverse section imaging because of the construction of the phantom. Instead, the extra thickness of activity that simulates a hot spot when viewed from in front appears as an isolated circle of activity on the corresponding transverse section.

Finally, the result of a preliminary quantitative study is summarized in figs. 9 and 10. A phantom was constructed of a hexagonal arrangement of small cylinders, each containing a different specific activity of ^{68}Ga (fig. 9a). The phantom was placed in the tomograph with the long axis of the cylinders parallel to the axis of rotation. The back-projection for an 8 mm thick section through the phantom is shown in fig. 9b for a pixel size of $2\text{ mm} \times 2\text{ mm}$ and 130,000 counts in

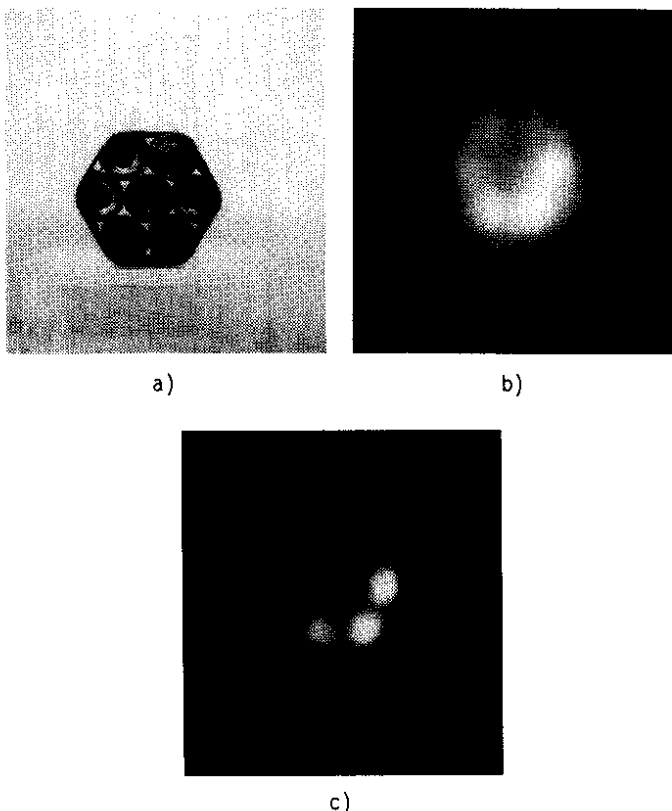


Fig. 9 Section through a quantitative test phantom: a) the phantom; b) back-projection; c) reconstruction.

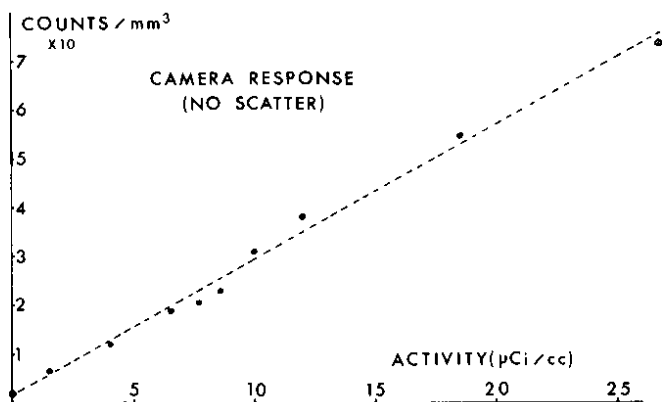


Fig. 10 Correlation of mean counts/unit volume with specific activity.

the image. By outlining each cylinder on the reconstruction (fig. 9c), the mean number of counts per unit volume can be computed and correlated with the known specific activity in the cylinder. For this example, the central cylinder contained only water, and for the other six the ratio of maximum activity to minimum activity was a factor of 5.7. The result of the correlation is summarized in fig. 10 for two separate data runs with different activities. Good linearity is observed over a wide range of specific activities.

These and other similar tests are being repeated in a scattering medium, but it seems clear that high-resolution image quantification should be possible with the rotating positron tomograph.

5. Conclusion

The above results have been presented to illustrate the success of the reconstruction algorithm, and to indicate the possibility of fully quantitative transverse section imaging with this high-resolution tomograph. However, many factors remain to be studied, such as the influence of scattered events on the final quantitative accuracy.

In view of the small imaging volume (700 cm^3) and long imaging times (13 min for 10^6 events) of the current prototype, larger ($30 \text{ cm} \times 30 \text{ cm}$) and more efficient (20%) HIDACs are at present under construction⁵. In addition, the new tomograph will have four detector units instead of only two, thereby increasing the data rate and reducing the number of rotational positions required. However, the algorithms presented and demonstrated above will form the basis of the image reconstruction software for the future system.

Acknowledgements

We would like to thank A. Wetherell, Experimental Physics Division, CERN, and P. Zanella, Data Handling Division, CERN, for their continuous support.

This work is supported by the Fonds national suisse, grant requests 3.849-0.79 and 3.986-0.80. R. Magnanini is supported under the request 3.986-0.80.

REFERENCES

1. A. Jeavons, B. Schorr, K. Kull, D. Townsend, P. Frey and A. Donath, Proc. Conf. on Medical Radionuclide Imaging, Heidelberg, 1980 (IAEA, Vienna, 1981), p. 49.
2. K.C. Tam, Ph.D. Thesis, report LBL-10039 (1979).
3. K.C. Tam and V. Perez-Mendez, report LBL-10207 (1979).
4. B. Schorr and D. Townsend, Phys. Med. Biol. **26**, 305 (1981).
5. A. Jeavons, K. Hood, G. Herlin, C. Parkman, D. Townsend, R. Magnanini, P. Frey and A. Donath, The high-density avalanche chamber for positron emission tomography, this Conference.
6. D. Townsend, R. Clack, P. Frey, A. Donath, A. Jeavons and J. Kulick, Data acquisition and image reconstruction for a high-resolution positron camera, in Proc. Conf. on Frontiers of Computers in Medicine, Philadelphia, 1982.
7. B. Schorr, D. Townsend and R. Clack, A general method for three-dimensional filter computation, Submitted to Phys. Med. Biol. (1982).
8. I.M. Gel'fond and G.E. Shilov, Generalized functions (Academic Press, New York and London, 1964), Vol. 1.
9. G. Chu and K.C. Tam, Phys. Med. Biol. **22**, 245 (1977).
10. J.G. Colsher, Phys. Med. Biol. **25**, 103 (1980).

