# Feasibility of Time-of-Flight Reconstruction in Positron Emission Tomography

Nizar A. Mullani, Joanne Markham, and Michel M. Ter-Pogossian

*Mallinckrodt Institute of Radiology and Washington University Medical School, St. Louis, Missouri*

**A feasibility study was carried out to determine whether image quality can be improved by the use of time-of-flight (TOF) information in positron emission tomography (PET). The experiment used two fast cesium fluoride detectors followed by constant-fraction discriminators for coincidence-timing resolutions of 600 to 800 psec full width at half maximum, depending on the energy discrimination level. A point source was scanned to study the spatial response of the point spread function with and without the TOF information for nonfiltered back-projected data. Back-projected images of a simplified chest phantom, 42 cm in diameter and filled with relative activity concentrations of 1,0, and 5, are presented for the unfiltered data to demonstrate the improvement in image quality obtained with the use of TOF. Filtered and reconstructed images of this phantom are also presented to show the relative differences in the images obtained with PET and TOF-PET techniques for similar filter functions.**

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Several researchers (*1,2*) have suggested using the time of flight of positron annihilation photons to determine the location of the positron. There have been two major drawbacks to this technique in the past: (a) lack of a suitable detector, and (b) lack of an appropriate mathematical algorithm. Allemand et al. (*3*) have shown by mathematical simulations that the combination of time-of-flight data with computed tomography should result in an improved signal-to-noise ratio in the reconstructed image. They also suggested the use of cesium fluoride (CsF) (*4*) as a scintillator to improve time-of-flight information.

We have used cesium fluoride detectors in our new PETT VI tomograph and have found them to be superior to sodium iodide (NaI) in detection efficiency and coincidence timing. We also find that because of the fast scintillation decay of this material, and the number of light photons generated within the first few nanoseconds of the scintillation, it is possible to obtain coincidence timing of 600 psec FWHM for two detectors. This coincidence timing is a measure of the time difference between the detection of the two annihilation photons at the two detectors, and can be used as a measure of the location of the source with an uncertainty in position of about 9 cm FWHM. Given this uncertainty, we wish to determine whether the incorporation of TOF information can result in improved image quality in PET reconstruction.

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For reprints contact: Nizar A. Mullani, MS, Div. of Cardiology, Univ. of Texas Medical School, P.O. Box 20708, Houston, TX 77025.

## METHOD AND RESULTS

In conventional PET systems using NaI or bismuth germanate (BGO) (*5*) and a coincidence timing window of 15–20 nsec, the time-of-flight information is not available, and therefore we assume that the positron annihilated somewhere within a cylindrical volume enclosing the two detectors. With fast detectors such as CsF and plastics (*6*), we can measure this time-of-flight distribution and locate with a known uncertainty the position of the annihilating positron. This added information should aid in placing the inferred origins closer to the true positions in an image reconstruction. To ascertain whether this is the case, we scanned a point source of positrons placed between two CF detectors that were set 100 cm apart. The detectors were CsF crystals

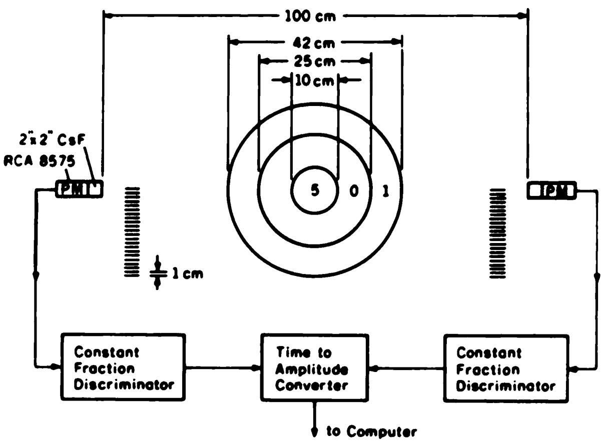


**FIG. 1.** Plan view of experimental setup of two CsF detectors and data-collection technique used for this study. Cylindrical phantom was filled with relative activity concentrations of 1, 0, and 5.

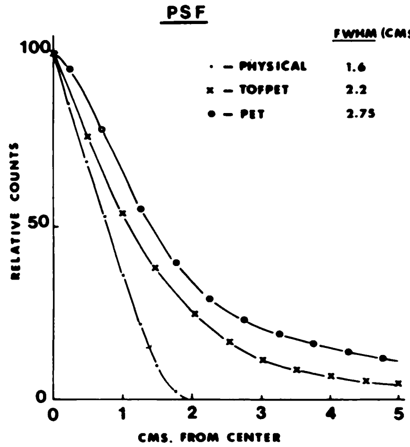


**FIG. 3.** Cross-sectional profiles of point source (point spread function) for nonfiltered PET and TOF-PET modes, compared with physical resolution. Because of symmetric nature of data, only one half of profile is shown.

5.08 cm in diameter and length (collimated down to 4 cm diameter opening), optically coupled to photomultiplier tubes and followed by constant-fraction discriminators set at 450 keV. The time difference between the two discriminator outputs is digitized and transferred to a computer as shown in Fig. 1.

The point source was scanned every 0.5 cm by moving it in a direction perpendicular to a line connecting the two detectors. At each scan position, a set of TOF data is collected by using the time difference between the detected photons to determine the longitudinal location (0.67-cm intervals) of each coincidence event along the line between the two detectors. As the source is moved away from the position of the axial line connecting the two detectors, the total number of detected events decreases due to the physical resolution of the detectors (7). The combined data represent a complex three-dimensional distribution of counts with the scan direction along one axis, the TOF distribution along the second axis and the intensity along the third. We show this in a two-dimensional isocontour format (Fig. 2) with the intensity normalized to 100. The measured physical resolution is 1.6 cm FWHM and the TOF uncertainty is 9 cm FWHM.

The data are then reconstructed on a 100- × 100-pixel array, without filtering, in the conventional mode using a back-projection method, and in the TOF mode by weighting the back projection with the time-of-flight information. The TOF-weighted back projection is equivalent to superimposition of the TOF data onto the array for each scan point, since the collected data are used as the weighting function along the back-projection line. Cross-sectional profiles of the reconstructed point source for the nonfiltered PET and TOF-PET images, and for the physical resolution of the two detectors, are shown in Fig. 3. All three curves have been normalized to the peak value to show the resulting spatial distribution of the point spread function before filtering, which is 2.75 cm FWHM for PET, 2.2 cm FWHM for TOF-PET, and 1.6 cm FWHM for the physical resolution.

To demonstrate the feasibility of TOF-PET in clinical situations, we constructed a chest phantom consisting of three concentric rings containing relative activity concentrations of 1, 0, and 5. The phantom as shown in Fig. 1 was scanned at 1-cm intervals with the two CsF detectors. The energy discriminators were set at approximately 125 keV, with a resulting time-of-flight

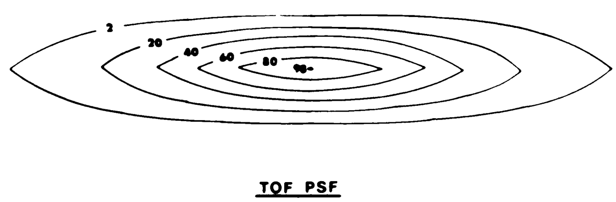


**FIG. 2.** Isocount contours of time-of-flight response function for a point source scanned between two CsF detectors, with physical resolution of 1.6 cm and TOF resolution of 9 cm (600 psec) FWHM.

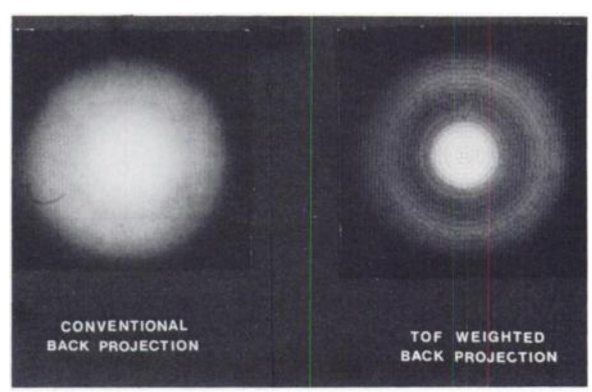


**FIG. 4.** Nonfiltered back-projected images of the 42-cm phantom, obtained by conventional back projection and TOF-weighted back-projection techniques as described in text.

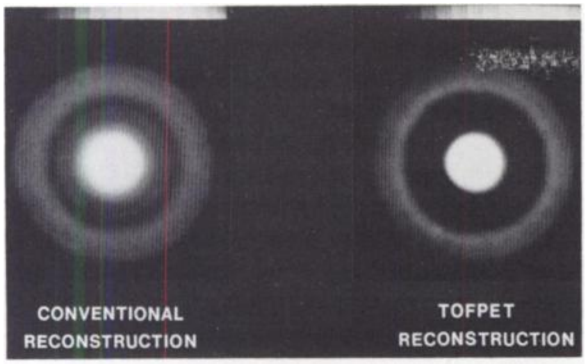


**FIG. 5.** Filtered reconstructed images of 42-cm phantom. Conventional PET image is obtained by filtered back-projection method, and TOF-PET image by two-dimensional deconvolution filtering of the unfiltered TOF image.

resolution of 800 psec (or 12 cm) FWHM. A scan profile comprising 22 scan points was collected as described earlier (with intervals of 0.83 cm for TOF data) with a total count of 33,000 coincidence events in the scan profiles. Because of the symmetrical nature of the phantom, the same scan data are used at 90 different angles for the reconstruction process with a 100- × 100-pixel reconstruction array and a 50-cm field of view.

To evaluate the image quality before filtering, the data were first back-projected on to the array in the same manner as for the point source. The two images thus obtained in the PET and TOF-PET modes are shown in Fig. 4, which indicate that a significant improvement in image quality can result from the addition of the time-of-flight information. The data were then reconstructed using the filtered back-projection method for the conventional PET mode and a two-dimensional deconvolution of the TOF image for the TOF-PET mode. The frequency responses of the two filters used and the contrast setting for the two photographs were similar in both. Some of the artifacts as seen on the images are due to the sampling resolution and the propagation of the same noise at each angle from the use of a single-scan profile. The two filtered images are shown in Fig. 5, and the profiles through the centers of these images are shown in Fig. 6. Note that in these figures the noise and the spatial response of the TOF-PET image is better than in the conventional PET mode, for similar reconstruction filters.

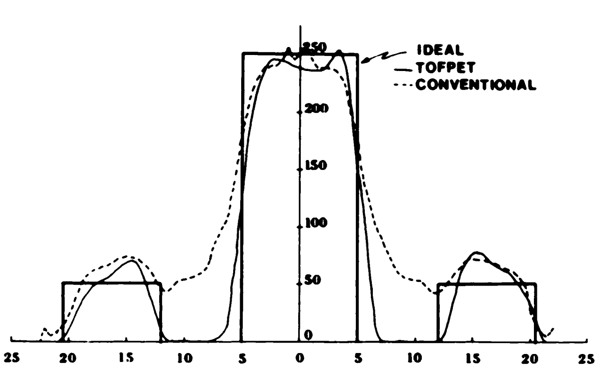


**FIG. 6.** Profiles of two images in Fig. 5, compared with ideal cross-sectional profile through center of phantom.

## DISCUSSION

We have shown the feasibility of incorporating the time-of-flight data in PET systems by using CsF detectors and reconstructing the TOF-PET images of a phantom representing the human chest. We have also shown that an improvement in image quality can result from this added information for both the back-projected and the filtered reconstruction images. No attempt has been made to quantify the improvement in signal-to-noise ratio in a TOF-PET image compared with a conventional PET image, since it would require data collection at all angles with multiple detectors. However, the results of this study indicate that the incorporation of TOF information in PET scanners should be a useful means of improving PET images. There still remain some technical challenges in implementing TOF-PET in a clinical situation, since a large number of detectors and very fast electronics are required. These technical difficulties can be overcome by the use of fast microprocessor-based electronics for on-line data processing. With the better CsF detectors and faster, small photomultiplier tubes available today, construction of a TOF-PET scanner becomes practical as well as feasible.

## ACKNOWLEDGMENT

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