Positron Ring System Using Anger-type Detectors

Progress Report

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Abstract

The major accomplishments of this year include 1) building and debugging a new set of coincidence electronics for our laboratory setup, 2) performing a series of detector experiments in the dry glove-box aimed at improving the performance of NaI(Tl) position-sensitive detectors, 3) modifying and debugging a Monte Carlo simulation code to test reconstruction algorithms and predict overall performance of a large solid angle PET scanner, 4) significant progress in the 3-D reprojection reconstruction algorithm and comparison to the 2-D single-slice algorithm and a 3-D multi-slice rebinning algorithm, 5) performance comparisons of the two PENN-PET scanners, which lead to a design for a large solid angle scanner with a 25-cm axial extent.

Technical Progress Report

One of the major goals of this 3-year project is to extend the design of the PENN-PET scanner to a one with a much larger axial field-of-view. We hope to improve the sensitivity, spatial resolution, and countrate capability of the system, while retaining the advantages of continuous position-sensitive detectors. Before building the new system, we have pursued experimentation and computer simulations aimed at helping decide on the parameters and configuration of the new system. At the same time, we have continued working on a 3-D reconstruction algorithm, since this will be required for a system with a large axial acceptance angle.

1. Electronics

This year we tested and debugged a new set of electronics for the laboratory bench set-up. These electronics are based on the newer design from UGM Medical Systems, which replaces the original version developed at the University of Pennsylvania. The electronics includes 64 channels of preamplifiers and digitizer/integrators, as well as a position calculator, master controller board, and single board computer to communicate with the SUN Sparcstation 330. The calculator has not yet been tested, and so the data are currently analyzed on the SUN computer, which is adequate for low countrate, spatial resolution tests. These electronics are much more flexible than the older set, and will enable us to collect data from a whole system of detectors and multiple coincident pairs, which we could not do with the old set. The new calculator will enable us to evaluate local coincident triggering, which allows us to sub-divide the large detector into smaller units, and therefore improve the high countrate capability of the system.

2. Detector experiments

Experiments were performed to improve the performance of NaI(Tl) position-sensitive detectors. Most of these experiments focused on the spatial resolution and edge effects of the detectors, and are described in [1]. While we have reduced the effects of the gaps using a software correction technique in the current PENN-PET, which uses six large detectors, we believe that there is room for improvement in the detector itself.

Many of the experiments on the detectors required the use of the dry glove-box, which allowed us to work with unencapsulated crystals. We investigated 1) optically coupling neighboring detectors directly to each other and 2) surface treatment of the edges and front face of the crystal to improve overall light collection and hence position resolution and to increase the usable area of the crystal, thus reducing the gaps. Coupling two discrete detectors to each other is feasible and does yield appreciable light transmission across the gap, however the light spreads nearly uniformly to the adjacent detector, which yields little position information using the current weighted centroid algorithm. Better results would be expected with a coupling compound of the same index of refraction as NaI(Tl) (n = 1.85), which is not available. The surface treatments included modifying the edge reflections with a black absorber, and machining grooves in the front surface to narrow the light spread in the crystal. This technique was shown to improve the overall spatial resolution by 20%, and could possibly be modified to preferentially improve the performance at the edge of the detector. Additionally, we investigated different photomultiplier tube (PMT) configurations to increase the sampling of the light at the edges. By using a thin PMT at the edge, the unusable area was decreased by 50%. This would translate to a reduction of the gaps between detectors in the PENN-PET scanner from 5^{o} to 2.5^{o}.

During the year we gained a lot of experience in handling unencapsulated crystals and learned how to modify and test the crystals. Some of this knowledge will be incorporated into the design of the new detector for the large solid angle scanner. In addition, we will continue to work on ideas to improve the position-sensitive detector.

3. Computer simulations

This year we have developed a Monte Carlo simulation code for the PENN-PET scanner geometry. Based upon earlier work at the University of Pennsylvania, the code can simulate list-mode data for a variety of objects, which can then be reconstructed using our reconstruction algorithms. In addition, the geometry of the scanner can be changed easily, to allow us to vary the diameter and axial extent of the scanner.

The code can simulate the ideal case without attenuation and scatter, or it can simulate the more realistic case with these physical effects. In the ideal case, the simulated data serves as a way to test and compare different reconstruction algorithms, specifically the 2-D vs. 3-D algorithms. With the physical effects included, the simulated data allows us to predict the behavior of the system as the diameter and axial extent are varied.

4. 3-D reconstruction algorithms

In volume imaging without septa, it is natural to move from a 2-D reconstruction algorithm to a 3-D algorithm, particularly as the axial acceptance increases. We continue to be concerned, however, with the practical implementation and computer time requirements of a 3-D algorithm. In addition, the 3-D algorithm needs to incorporate new correction methods for quantitation, including normalization, scatter correction, attenuation correction, and randoms correction. We have therefore evaluated the spatial resolution and image quality of the PENN-PET, using 2-D and 3-D reconstruction algorithms.

The data from the PENN-PET are normally rebinned (in hardware) into two-dimensional sinograms (256 rays x 192 angles), whose slice number is determined by averaging the axial coordinates of the two coincident detectors. This single-slice rebinning method is a geometrical approximation for oblique rays, except for those events originating at the center of the scanner. The slices (up to 64 for the UGM scanner) are then reconstructed independently by filtered backprojection, with corrections for efficiency normalization, scattered and random coincidences, attenuation, and gap compensation. In addition, an axial normalization is performed to compensate for the non-uniform axial sensitivity.

We have continued to develop a fully three-dimensional algorithm [2], which requires the data to be spatially invariant. This is accomplished by estimating missing data by reprojecting through a two-dimensional reconstruction of the object. The missing data are due to the finite axial extent of the scanner, which occurs with every practical scanner, as well as the gaps between the detectors, which is a characteristic of the PENN-PET. While this algorithm is more accurate for oblique rays, it requires more computation and different methods of quantitative corrections.

The first step towards our implementation of the 3D reprojection reconstruction algorithm was the porting of the programs we used for our initial feasibility study from a VAX/VMS environment to a SUN/UNIX environment. After repeating our feasibility tests in the new computer/operating system environment we increased the running speed of the image reconstruction process by using a new algorithm for the 3D forward- and back-projection steps, which is an extension of a 2D algorithm. This resulted changing the numerical integration along the a line-of-sight from trilinear interpolation to a stepped bilinear interpolation without any loss in accuracy and with a considerable gain in speed. The routines were then carefully hand-optimized and debugged. These two procedures lead to a decrease in execution time from 29 hours to 4-1/2 hours. It is anticipated that when these routines are implemented on an array processor, the execution time will drop to under an hour.

During the optimization of the image reconstruction process, we also designed, tested, and implemented algorithms necessary for a practical implementation, namely (1) Compensation for the transverse detector gaps. This was done by reprojection through the detector gaps at the same time the reprojection was done to compensate for the finite axial extent of the scanner. This method was shown to work, however it requires several iterations, and it seems likely that reprojection in 2D would be more efficient. (2) Attenuation correction. This was done by using a straight forward geometrical correction that takes into account the size and position of the phantom being scanned. This worked well, but a more robust method will be needed for patient scanning. (3)

Corrections for random and scattered coincidences. We extended the method that we currently use in 2D reconstruction to 3 dimensions to correct for random and scattered coincidences. This method performs well with the phantoms used to date, however a more sophisticated method, such as dual- or multiple-energy window method will likely be required for patient studies.

A less critical, but still necessary correction that we are still testing is the correction for detector nonuniformities. We have developed a technique that is similar to a 3D extension of the standard 2D method that is applicable to the 4D data set that we use. It works by first computing the averaged detector channel sensitivities and then calculates the product of the sensitivities (for a coincident given event) by a 3D coordinate transformation.

A different three-dimensional algorithm has also been tested, multi-slice rebinning [3], which was developed by Dr. Robert Lewitt of the University of Pennsylvania while on sabbatical at UGM Medical Systems. This algorithm attempts to achieve a higher degree of geometrical accuracy for oblique rays without the computational burden of a fully three-dimensional algorithm. This approach rebins oblique rays into multiple sinograms, depending on how many are intersected by the coincident line. The data are then reconstructed two-dimensionally, with the same quantitative correction methods as those applied above. This step is immediately preceded or followed by axial filtering to reduce the blurring in the axial direction, which is independent of the filtering in the transverse direction during reconstruction.

Neither 3-D algorithm (the multi-slice rebinning and 3D-reprojection) is implemented in hardware yet, and so the data to test them must be taken in list mode. While this is useful to test the algorithms with point source data and a limited number of phantoms, it is not practical to acquire list mode for a patient study, because of the limited disk transfer time and disk space. We will continue with simulations to more carefully evaluate the algorithms, particularly for a scanner with a much large axial extent, as described below.

5. Large solid angle PET scanner

In order to understand the tradeoffs in volume imaging without septa, we have evaluated and compared the performance of our two PENN-PET scanners [4]; the first is the proto-type built at the University of Pennsylvania with a 10-cm axial extent and 9-cm axial field-of-view (FOV), and the second is from UGM Medical Systems with a 14-cm axial extent and 12.8 axial FOV. The scanners were compared in terms of spatial resolution, scatter fraction, sensitivity, countrate capability, and image quality. An increase in the axial length of the detector from 10 cm to 14 cm leads to an increase in system sensitivity of about a factor of two, from 65 to 130 kCPS/μCi/cc, with 16 kCPS/μCi/cc/axial-cm in the center. This is achieved with a maximum axial acceptance angle of $\alpha = 6.5^{\circ}$, which is the limit of the proto-type scanner in the center, and results in a more uniform sensitivity for the central slices in the UGM scanner. The main reason for limiting the acceptance angle with the UGM scanner is to limit the degradation in spatial resolution at large radii when using a two-dimensional reconstruction algorithm. With a three-dimensional algorithm, the acceptance angle can be increased to $\alpha = 9.0^{o}$ in order to increase the sensitivity near the center to 24 kCPS/µCi/cc/axial-cm. In addition, the spatial resolution is much more uniform, 5 to 7 mm, out to a radius of 10 cm.

With the larger axial FOV, the scatter does not increase for a head sized phantom. For an energy threshold of 450 keV, the scatter fraction is 13%. In addition, neither the scatter or random fractions change significantly with slice location, since both the scatter and randoms have an axial profile similar to the true axial profile. The random fraction was shown to decrease significantly, as a function of true countrate, for the larger UGM scanner. Therefore, for a given activity, the larger UGM scanner will produce a higher true countrate than the proto-type scanner, because of the higher sensitivity, with a lower ratio of randoms to trues. Alternatively, a lower activity level with the UGM scanner can be used to produce the same true countrate, also with a lower ratio of randoms to trues. The high countrate limitations, caused by electronic deadtime, are currently being investigated.

Since we have shown the larger scanner to offer significant advantages, we have decided to extend the axial extent of the scanner even further, to 25 cm. An advantage of the PENN-PET system, with continuous detectors, is that both the cost and complexity increase slower than the performance increase as a function of the axial FOV. For example, the UGM scanner offers almost 50% more axial FOV and 100% higher sensitivity than the proto-type scanner, with 50% larger detectors and only a 33% increase in the number of PMTs. If we restrict our design to brain imaging, an axial FOV of 12.8 cm is large enough, and so an increase in the axial size of the detector will serve primarily to increase the sensitivity. By focusing on the brain, we do not have to be overly concerned about the scatter or random fractions increasing, since activity in the body will still be outside the FOV. With an increase of the size in the axial dimension to 25 cm, and a decrease in the diameter to 42 cm, the sensitivity will improve by a factor of ten over the proto-type scanner, for a head-sized object. In fact, the sensitivity at the edge of the brain would be higher than that of the proto-type scanner at the center of the brain.

We are planning to build this scanner as a cylindrical system, using a single detector. The edge effects of the PENN-PET detectors has always been an important consideration, although the constrained Fourier gap compensation technique has allowed us to keep the system stationary during data acquisition. After considering the technical and practical constraints of eliminating the edge effects of position-sensitive detectors [1], we decided that we would achieve the best results with a cylindrical detector, which has no edge effects in the transverse direction. However, we still wish to maximize the axial extent by reducing the edge effects in the axial direction. In addition, with a local coincidence triggering system, it is now possible to electronically divide a continuous detector into separate channels, thus identifying coincidences in a single detector.

The detector will be 1.9-cm thick, with a 1.25-cm thick light guide. The thinner crystal, compared to the 2.5-cm thick PENN-PET detectors, is expected to be about 17% less efficient, leading to 33% less coincident detection efficiency. However, the tremendous gain in geometrical efficiency will make up for this loss. In addition,we expect a significant improvement in the spatial resolution,since the thinner crystal will have less spatial resolution degradation due to Compton scattering and the spread of the light distribution. The crystal will be coupled to thirty columns and six rows of 5-cm square PMTs, for a total of 180 PMTs. With a PMT channel width of 5 and overlap of 2 in the transverse direction, the total number of zones would be 10 for the system, with a total of 25 possible coincident pairs. This would ensure a good compromise to maintain detection uniformity and avoid excessive triggering of multiple channels.

Plans for 1992

For next year we plan to start testing the cylindrical scanner, which has already been ordered. We have enough photomultiplier tubes and electronic channels to perform tests on the spatial resolution and countrate capability. With the new position calculator and master controller board, we will evaluate the zone electronics to optimize the size of the local coincident zone, in terms of countrate capability. We will also set up a rotating platter and circular lead mask to implement a method of spatial linearity correction (distortion removal).

Work will continue on the 3-D reprojection reconstruction algorithm. The projects include (1) the testing and implementation of the procedure for the correction of detector non-uniformity and (2) the porting of the reconstruction programs to an array processor to further reduce the executing time to the range of 1-2 hours. This last step will be delayed until it seems likely that no further modifications to the reconstruction code seem likely. In addition there are several projects will likely be necessary after our testing of the algorithms with a more realistic brain phantom, in particular (1) using an attenuation correction algorithm for irregular objects and (2) a background correction method that uses a dual- or multiple-energy window method. The final step in this process will be the correlation of the result of measured and simulated data for the two scanners (with different axial field-of-views) by comparing the results of using 2D and 3D reconstruction algorithms for different phantoms. These results will be used in conjunction with Monte Carlo simulations to extrapolate the performance of a very large axial field-of-view system.

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