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DATA ACQUISITION, RECONSTRUCTION AND DISPLAY FOR THE DONNER 280-CRYSTAL POSITRON TOMOGRAPH

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ABSTRACT

The imaging system for the Donner 280-Crystal Positron Tomograph is described. The architecture of the hardwired data acquisition system (histogrammer) allows a data rate in excess of two million events/second while simultaneously correcting for accidental events. The histogrammer has eight memory cards to allow separate accumulation of time-slices of the cardiac cycle, or multiple-buffering for rapid sequential studies. Transfer time from the double-bussed histogrammer to a disk of the host computer system is less than one second. A hardwired device is able to reconstruct a 256x256 image from 140 angles in less than three seconds, but since calibration and correction for attenuation are presently performed in software, this time is increased to about 10 seconds. Images are displayed on a 256x256 raster scan display system with image manipulation capabilities.

INTRODUCTION

This paper describes the data acquisition, reconstruction and display system for the Donner 280-Crystal Positron Tomograph[1,2] which is a circular ring system for transverse section tomography using annihilation photons from positron emitting radiopharmaceuticals. Such systems are gaining wide acceptance at this time for the quantitation of the amount of isotope in well defined regions of the body resulting in the deduction of functional or metabolic activity without discomfort to the patient. The architecture of the data acquisition, reconstruction and display system described here is general in nature and is applicable to many circular ring systems.

For reviews of instrumentation in positron computed tomography see references 3 and 4. For a guide to the practical application of emission computed tomography see reference 5.

OVERALL CONTROL

Overall control of the data acquisition, reconstruction and display system is handled by a DEC PDP-11/10 computer. The PDP-11/10 computer has an architecture which permits the interfacing of user designed peripherals in a straightforward manner. In addition, the architecture is identical to the rest of the PDP-11 family of computers, enabling the user to upgrade the central processor with no change to peripheral hardware.

The computer performs routing of data between devices peripheral to the central processor, and it is intended that a minimum amount of numerical computation be handled by computer software. Figure 1 is a diagram showing the interconnections between the central processor and the peripheral devices which presently make up the system.

DATA ACQUISITION

The data acquisition system for the Donner 280-Crystal Positron Tomograph is a hardwired device which accumulates data passed to it from the ring coincidence electronics. It contains its own control logic, address generation logic and histogram memory. Control functions of the device are performed by an M6800 microprocessor. The M6800 responds to commands from the computer and initializes the histogram memory, starts and stops data accumulation and initiates transfer of data to the PDP-11 interface.

The data from the ring are in the form of two crystal addresses and an on-time/off-time flag. The chord passing through the two crystals is required to pass through the 50 cm. patient port, and the two crystal addresses can be transformed into the chord's angle and distance from the center of the patient port. Legal events can then be described by one of 280 angles (between zero and π) and one of 105 (positive and negative) lateral distances. This corresponds to each of the 280 crystals being put into coincidence with the opposing 105. If the data are organized in this fashion, we have 280 projections, but each projection has alternately 53 and 52 data points (for odd and even projection numbers, respectively). For purposes of calculating consecutive memory addresses to histogram the data, we merge even and odd projections to form 140 projections each with 105 data points. Within each projection, half of the data points differ by an angle of $\pi/280$ with the other half.

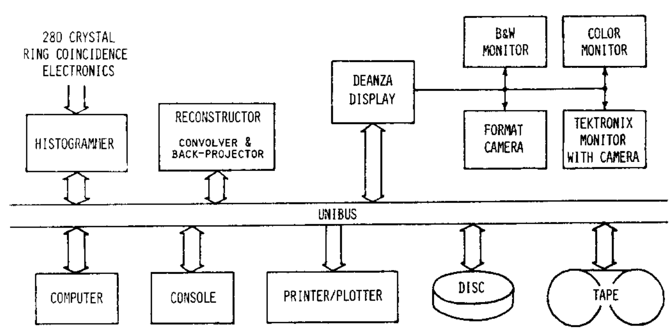

Figure 1. Schematic diagram of the data acquisition, reconstruction and display system for the Donner 280-Crystal Positron Tomograph.

Responding to an on-time/off-time flag, the contents of the correct histogram memory word are either incremented or decremented. Therefore, at the end of data collection, the contents of each histogram memory word represent true coincidences corrected for accidental coincidences which have occurred. Within the ring electronics, in addition to the normal coincidence circuits, there are circuits which test a delayed pulse from each of the 280 crystals with the prompt pulses of the opposing 105 for coincidence. If such a coincidence occurs it must be purely by accident since one of the pulses has been delayed. Such events occur to an equal extent in the "on-time" data, the two photons having originated from different positrons and having been detected within the coincidence resolving time by chance. The accidental events in the "on-time" data are accounted for by incrementing the histogram memory by one for each "on-time" event and decrementing the memory for each "off-time" event. In this way real time accidental subtraction is performed while the data are being accumulated. The variance of a resulting data point is not the contents of that histogram bin, but is instead given by the sum of the corresponding "on-time" and "off-time" events.

A block diagram of the data acquisition system is shown in figure 2. The memory address calculation takes about 400 nanoseconds and does not interfere with the memory timing. The accumulation of one event into histogram memory can proceed while the memory address calculation for the next event is taking place. An array of 13x15 1K word by one bit CMOS memory chips are contained on each histogram memory board. The read-increment-write cycle time for the memory chips used is about one microsecond and is sufficiently fast for the present system. By using the same architecture and faster memory chips, the histogrammer speed could be increased by more than a factor of two.

The histogram memory consists of eight 105x140 = 14,700 word by 12-bit sections (not including patient identification data, imaging parameters and parity). These eight separate memories are required in order to stop heart motion for gated cardiac studies or in order to collect data for many sequential studies without interruption.

In the gated mode, the M6800 microprocessor responds to an EKG signal from the patient as depicted in figure 3. The histogrammer contains eight memory boards in order to handle gated cardiac studies. Data collection starts on the R-wave of an EKG signal and for some preset time interval, usually 100 milliseconds, all events are accumulated in memory board 1; for the second time interval all events are accumulated in memory board 2. The process continues until all eight memory boards have sequenced or until a new R-wave re-initiates the procedure. It is assumed that the heart returns to the same physical position at the end of each cardiac cycle and that the motion is the same during each cycle. By gating data acquisition, we reduce motion of the heart to only that which takes place within a fraction of the full cycle. To accumulate sufficient data for statistically significant images, data are collected over many cardiac cycles (usually 5 to 10 minutes).

In order to accumulate data for many sequential images without interruption, the histogrammer was designed with a double-bussed structure as shown in figure 2. While data are accumulated into one histogram memory board using bus 2, data from a previous study can be simultaneously transferred out of any other memory board using bus 1.

DATA STORAGE

Short term storage of histogrammed data is provided by a DEC RK05 disk system with removable disk cartridges. Modification of the standard DEC I/O device driver for the RK05 disk allows us to transfer data directly from the histogrammer to the disk. One of the 15K word histogram memories can be transferred to the disk in about 1/2 second. The disk files are handled by the DEC PDP-11/10 computer, and data from 80 histogram memory cards can be stored on a single disk cartridge. The double-bussed architecture of the histogrammer together with the speed of the disk

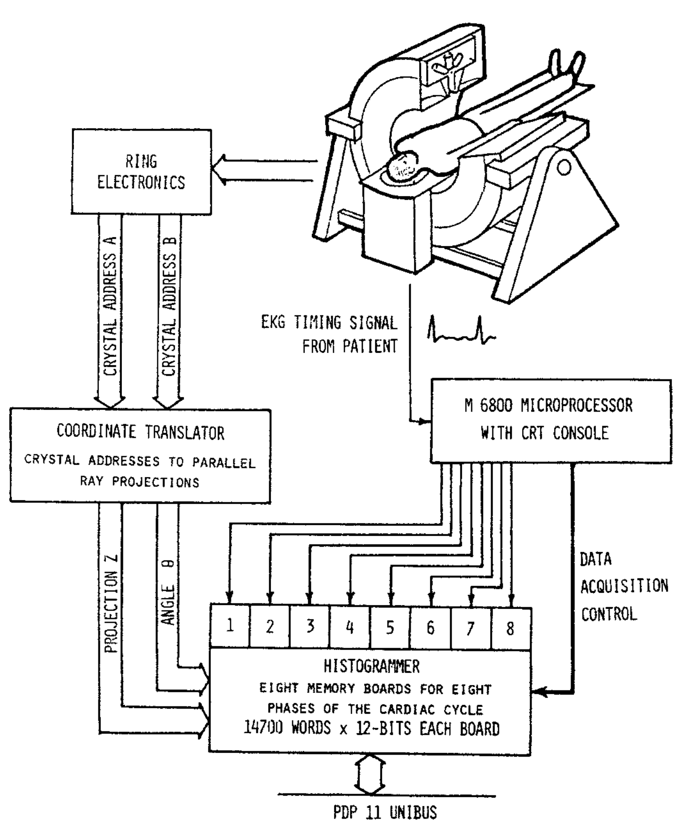

Figure 3. Control for gated cardiac imaging.

transfers permits us to take data for up to 80 sequential images as short as 1/2 second with out interruption.

Archival storage capability is provided by an IBM-729 tape drive. This 7-track drive operating at 800 BPI allows the storage of about 250 histogram memories on a 2400 foot magnetic tape.

DATA CORRECTION

Raw emission data from the histogram memories must be corrected for several sources of systematic errors before reconstruction into transverse section images can proceed. These effects are categorized as crystal-pair or chord inefficiency and attenuation due to intervening tissue.

Crystal pair inefficiency is due to factors such as crystal light output variability, photomultiplier tube variability, and coincidence timing inaccuracy. Correction for these effects can be performed by collecting a large amount of data from a known emission source and comparing the resulting data with theoretically known expected values. The ratio of theoretical to measured values yields a set of chord efficiency correction factors which are then applied to all subsequent histogram data.

The emission source we use to compute chord efficiencies is a hoop source of positron emitting isotope. The hoop is larger than the patient port and is placed in the plane of the ring, so that all 14,700 chords are excited. The hoop is cylindrically symmetric, so that all chords equidistant from the center of the patient port would be equally populated if the chord efficiencies were equal. We obtain chord efficiencies by taking the ratio of events collected for each chord divided by the mean number of events collected from all chords which are the same distance

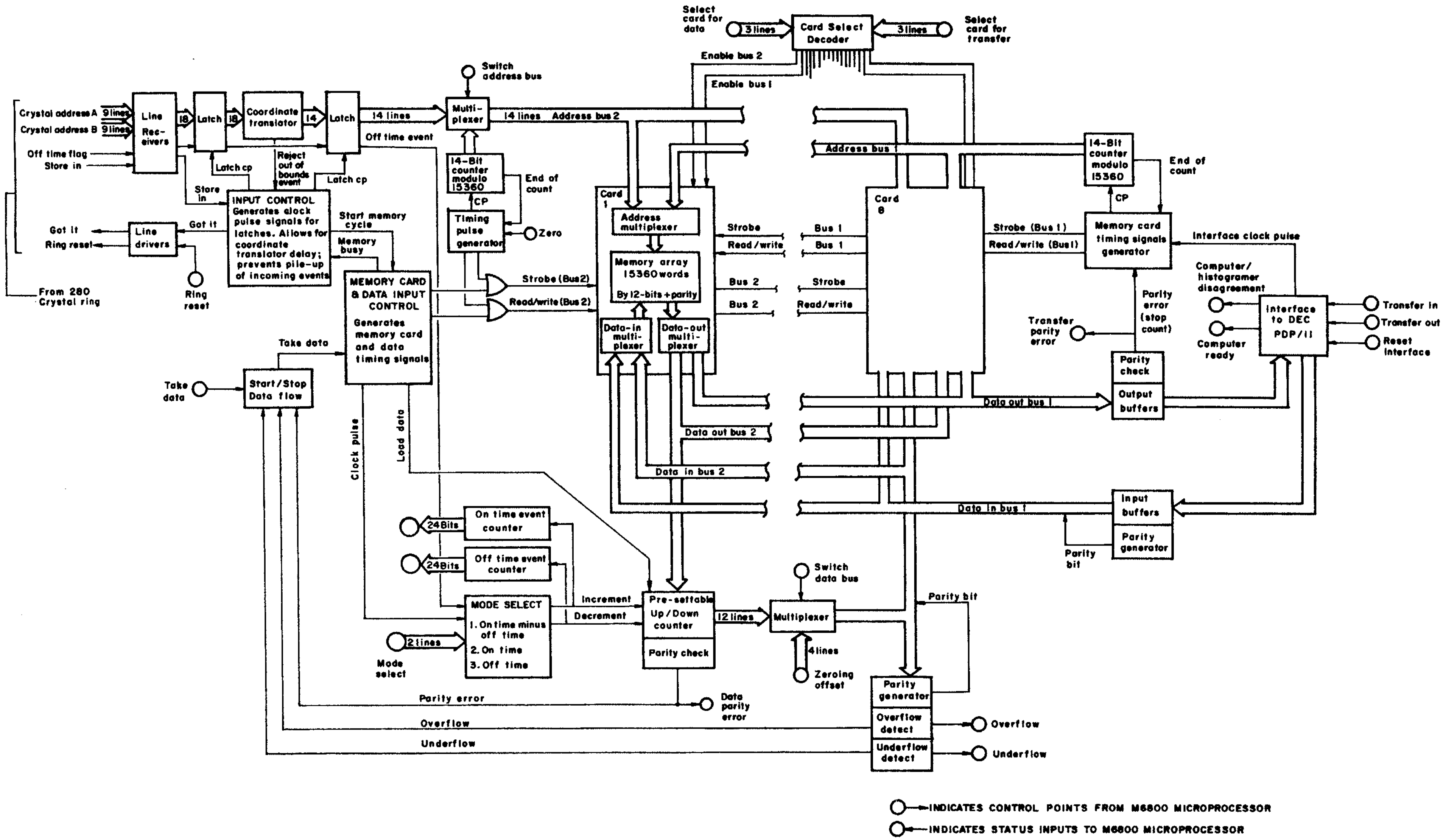

Figure 2. Block diagram of the data acquisition system.

from the center of the patient port. This simple procedure yields good results when the variability of chord efficiency is not too great.

Attenuation correction factors are calculated by collecting histogram data with the patient and an emission source (external to the patient) in the positron camera. The hoop source described above is used for this purpose. These data are compared to data collected with only the emission source present. The ratio of data collected with the patient in to data with the patient out is a measure of the attenuation of annihilation photons by the patient. The reciprocals of these values are the attenuation correction factors and are applied to all subsequent patient data (with injected radiopharmaceutical) which are accumulated with the patient in this position.

Because of the thickness of the human body in some sections, the transmission of 511 keV annihilation photons can be as low as 5%. This reduces the statistical quality of the transmission data and therefore reduces the accuracy of the attenuation correction factors. To help suppress these statistical fluctuations, we smooth the attenuation factors with a three-point filter (with values .25, .50, .25). The filter may be applied several times for sections with greater attenuation. After the attenuation factors have been filtered, their reciprocals are taken as the attenuation correction factors.

The combined chord inefficiency and attenuation correction factors are calculated by the PDP-11/10 computer. Correction factors are stored on the RK05 disk, and these correction factor files are utilized to correct subsequent emission data for the patient at the corresponding position.

IMAGE RECONSTRUCTION

Reconstruction of transverse section images of radiopharmaceutical concentration is accomplished using the convolution algorithm[6]. Data from each projection angle are convolved with a user-defined kernel and subsequently back-projected onto the image of the transverse section.

The convolution algorithm assumes that the projection data are collected at equally spaced lateral distances for each of a number of equally spaced angles. In positron ring systems, we are dealing with projections taken between crystal detectors which are equally spaced around the circumference of a circle. For the chords taken at a single angle, those further from the center of the patient port tend to be closer together. This effect is minimal for the Donner 280-Crystal Positron Tomograph for which the patient port is 50 cm. in diameter and the ring of crystals is 90 cm. in diameter. As was pointed out in the data acquisition section above, we have organized the histogram data into 140 projections with 105 data points each, whereas the data actually arise from 280 distinct angles. We have found no noticeable difference in the resulting images when these data are assumed to be taken at 280 or 140 angles.

Reconstruction is performed by an Analogic[7] AN-7510-I Digital Filter (convolver) and AN-7545-I/M Data Distributer (back-projector). This hardwired device reconstructs a 256x256 pixel transverse section from the 105x140 corrected data points in about 2.5 seconds. Because the convolver and back-projector act independently, convolution of data for one angle can take place simultaneously with the back-projection of convolved data for the previous angle. The back-projector has a "roam and zoom" capability, so that the image magnification and the center of the image with respect to the center of the patient port may be chosen by the user.

The total reconstruction time is presently about 10 seconds because the data correction is performed by the PDP-11/10 computer without hardware multiply capability. The back-projection is accumulated in a 256x256 word memory with 24 bits per word. This large word size minimizes the loss of accuracy due to round-off errors.

While the central goal of positron ring systems is the imaging of transverse sections of the distribution of positron emitting isotopes, there is a secondary by-product of the attenuation correction procedure. The logarithm of the attenuation correction factors are projections of the linear attenuation coefficient through that section of the patient's body. The linear attenuation coefficient is approximately proportional to density, and we routinely reconstruct transmission images which provide useful anatomical landmarks and registration for the corresponding emission images. Figure 4 shows a typical reconstruction of the distribution of a positron emitting radiopharmaceutical in the heart muscle of a dog. Figure 5 shows the corresponding transmission image.

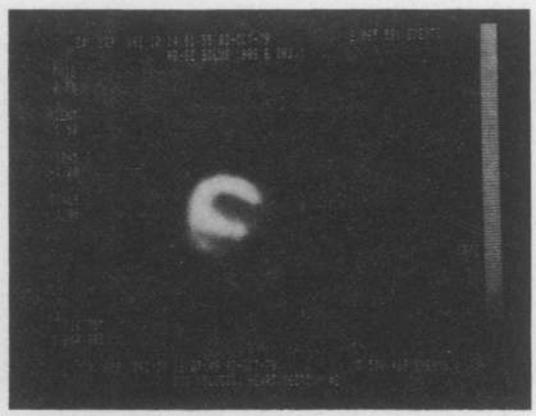

Figure 4. Emission image of a transverse section of a dog thorax including the heart.

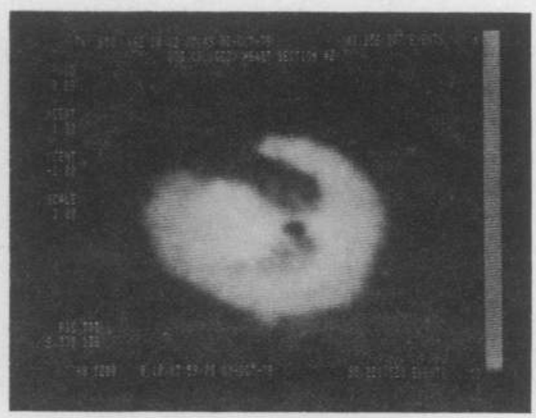

Figure 5. Transmission image of the transverse section of a dog thorax corresponding to the emission image of figure 4.

IMAGE DISPLAY

Reconstructed transverse section images are displayed using a DeAnza[8] ID-2212 Image Display System. The image display system has a 256x256 by 12-bit refresh memory. The upper two bits are utilized for overlay functions, so that ten bits are available for the reconstructed image. The video output provides an RGB interface for color monitors as well as normal and reversed monochrome monitor interfaces. An intensity transformation table maps the lower ten bits of refresh memory to the video output and allows windowing and other image manipulation to be performed. A monochrome monitor and a high resolution color monitor are used to view the final images. Single photographs are obtained from a separate monitor/camera configuration, and photographs of multiple images are obtained from a format camera.

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REFERENCES

1. Derenzo, S.E., T.F. Budinger, J.L. Cahoon, R.H. Huesman, H.G. Jackson, "High Resolution Computed Tomography of Positron Emitters", IEEE Trans. Nucl. Sci. NS24(1) pp. 544-588, 1977.

2. Derenzo, S.E., T.F. Budinger, J.L. Cahoon, W.L. Greenberg, R.H. Huesman, T. Vuletich, "The Donner 280-Crystal High Resolution Positron Tomograph," IEEE Trans. Nucl. Sci. NS26(2) pp. 2790-2793, 1979.

3. Derenzo, S.E., "Positron Ring Cameras for Emission-Computed Tomography," IEEE Trans. Nucl. Sci. NS24(2) pp. 881-885, 1977.

4. Budinger, T.F., S.E. Derenzo, G.T.Gullberg, R.H. Huesman, "Trends and Prospects for Circular Ring Positron Cameras," IEEE Trans. Nucl. Sci. NS26(2) pp. 2742-2745, 1979.

5. Huesman, R.H., G.T. Gullberg, W.L. Greenberg, T.F. Budinger, "RECLBL Users Manual - Donner Algorithms for Reconstruction Tomography," Lawrence Berkeley Laboratory publication PUB-214, 285 pages, 1977.

6. Ramachandran, G.N., A.V. Lakshminarayanan, "Three-Dimensional Reconstruction from Radiographs and Electron Micrographs: Application of Convolutions instead of Fourier Transforms," Proc. Nat. Acad. Sci. USA 68 pp. 2236-2240, 1977.

7. Analogic Corporation, Audubon Road, Wakefield, Mass. 01880.

8. DeAnza Systems, Inc., 118 Charcot, San Jose, Calif. 95131.