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Computers in Diagnostic Nuclear Medicine Imaging – A Review

K.K. Kapoor, R. Prakash, S. Khushu and R.V. Narayanan

Institute of Nuclear Medicine and Allied Sciences, Delhi–110 007

ABSTRACT

Digital computers are becoming increasingly popular for a variety of purposes in nuclear medicine. They are particularly useful in the areas of nuclear imaging and gamma camera image processing, radionuclide inventory and patient record keeping. By far the most important use of the digital computer is in array processors which are commonly available with emission computed systems for fast reconstruction of images in transverse, coronal and sagittal views, particularly when the data to be handled is enormous and involves filtration and correction processes. The addition of array processors to computer systems has helped the clinicians in improving diagnostic nuclear medicine imaging capability. This paper reviews briefly the role of computers in the field of nuclear medicine imaging.

1. INTRODUCTION

Digital computers are finding wide-spread application for a broad spectrum of purposes in nuclear medicine. Most of these applications make use of the ability of digital computers to perform complex operations on vast amounts of data in relatively short time. Computers are particularly useful in the areas of scan and gamma camera image processing. They are also used for analysing dynamic function data, and for general administrative functions, such as radionuclide inventory and patient record keeping.

For many years there has been an active interest in digital processing of scintigraphic images in nuclear medicine. Early interest was primarily in rectilinear scan image enhancement, whereas more recent efforts have been directed towards the collection and processing of data from scintillation cameras and emission

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computed tomography (ECT) cameras. Digital processing is useful for several purposes in a gamma camera study, including correction of image distortions arising from non-uniform or nonlinear detector response, improvement of image interpretability by smoothing or resolution recovery and quantitative analysis of sequential images in a dynamic function study.

1.1 Basic Principles and General Configuration

Digital imaging is an application tailor-made for minicomputers for use in nuclear medicine imaging. Basically, an electrical image produced from a gamma camera is converted into a computer-compatible form. Once the image is in this form, the minicomputer can process the data and provide meaningful results for the operator The conversion to computer-compatible format is achieved by using analog-to-digital convertor (ADC) circuits in the gamma camera interface. Usually two ADCs are used, one for the x-position and one for y-position signals, which are activated only upon receiving a z-signal from the camera indicating that the detected event was within the selected energy range.

The essential components of a typical scintigraphic data processor in a clinical setting include a minicomputer, a mass storage system, a data acquisition interface, a display generation module, a control console and a software system.

The first five components are hardware components. With the exception of the data acquisition interface and the display generation module, these are standard computer items such as the central processing unit (CPU), computer mainframe memory, peripheral storage devices and input-output devices. In a commercial system, the manufacturer is usually responsible for all components, hardware and software. The exact choice of individual hardware components varies widely among different manufacturers. Mass storage devices include magnetic disks and magnetic tape. Without exception, both are available with all systems, for both short- and long-term data storage. The data acquisition interface serves to translate signals from various imaging devices into signals acceptable to the computer. A wide variety of data display systems are used, but frequently these include a cathode ray tube device which may be a black and white graphic display terminal or a colour monitor. The control console is usually a keyboard operated unit, for example, a teletypewriter. Occasionally the control console and display device are combined into a single cathode ray tube keyboard unit.

Software varies widely from one system to another, as do the hardware components. However, five common features do exist. These are: (a) routine collection of static images with computerised image processing, including scintillation camera non-uniformity correction, smoothing, foreground and background suppression, region of interest quantification, and contrast enhancement; (b) maintenance of files of pertinent patients and study identification alongwith imaging data; (c) recording of dynamic function studies (rapid sequential camera images) and generation of count rate versus time histograms, with emphasis on zero computer contribution to dead time; (d) image presentation for interpretation and region of interest selection; and (e) availability of higher level programming languages to allow functional expansion to meet new clinical needs as they arise.

An additional hardware component which is usually added to the nuclear medicine imaging system is a fast array processor for emission computed tomography applications.

One of the early nuclear medicine image processors was the 50/50 system, manufactured by the Nuclear Data Company. The system contained three basic parts: (a) the PDP-8 minicomputer, (b) the image display unit (with memory), and (c) the magnetic tape storage unit. The PDP-8 minicomputer was used to save or retrieve nuclear medicine images from magnetic tape, manipulate images, and place images on the display screen. The limited memory capacity and the speed of the PDP-8 precluded sophisticated analysis of the data. Also, the limited speed of the magnetic tape prevented acquisition of fast dynamic studies. The image display unit augmented the computer by providing many hard-wired features such as background subtraction, isometric display, image size control, and profile selection. These features were useful but limited in scope and were basically confined to enhancement of the image display only. Quantitative analysis of nuclear medicine studies could only be performed by the PDP-8 minicomputer, and even simple computations required substantial lengths of time. The structure of the 50/50 system, however, established design principles for image processing systems that followed.

The present day nuclear medicine computer systems have advanced significantly in their image acquisition, processing and analysis and display capabilities. One of the typical systems called GAMMA-11 from DEC, USA, in extensive use at many nuclear medicine departments today, has the following configuration: (a) PDP-11/34 CPU-based minicomputer with 256 kb main memory and RT-11 foreground/background operating system, (b) LA 36 DecWriter terminal, (c) dual RL01 disk drives with 5 Mb capacity each, (d) Conrac VSV01 high-resolution colour monitor, (e) VT 100 black and white display console with keyboard, (f) GAMMA-11 nuclear medicine applications software, and (g) joystick/lightpen for marking regions of interest on displayed images. The additional components for enhanced storage capacity include a 9-track 800 bpi magtape drive and RL02 floppy disk drive.

The foreground/background monitor of the above system allows two programs to operate simultaneously: a foreground program and a background program. The real-time function, namely the acquisition of data from gamma camera, is accomplished in the foreground, which generally has priority on system resources. Functions that do not have critical response time requirements, such as data processing and analysis or program development, are accomplished in the background.

During non-activity on the data collection job, the computer can execute another program in the background in a different memory section. The computer thus can effectively be used for analysis of previous studies while it is collecting data from scintillation camera and awaiting completion of the study. The one limitation of foreground/background system is that it cannot support two programs that spend a majority of the time calculating; hence, simultaneous collection and analysis are allowed but not two simultaneous analyses.

The GAMMA-11 software intended for nuclear medicine imaging with a scintillation camera can be classified in seven general categories as given below:

(i) Data acquisition programs which allow definition of acqisition parameters, immediately before the study is performed or in advance of the study;

(ii) Transformations are performed on image data before display on the viewing screen. Display programs which offer control over many of the translation parameters. Variation of these parameters results in image enhancement;

(iii) Static image enhancement programs which include image smoothing, contrast enhancement, profile selection, isometric projection and contour mapping;

(iv) Dynamic image enhancement programs which include multiple view display and movie mode display;

(v) Methods for region of interest which include lightpen, cursor lines, joystick and automatic flagging;

(vi) Curve generation and manipulation programs which generate dynamic flow curves which can be used to quantitate parameters of isotope flow through selected anatomical regions. Two mathematical techniques commonly used in curve analysis are curve integration and curve differentiation; and

(vii) Utility programs which assist the operator in system utilisation, record keeping and organisation of patient data on storage media.

2. CLINICAL APPLICATIONS

2.1 Image Filtering

Image processing, with the intent of improving display information, was one of the first applications of the computer in nuclear medicine. Early investigators hoped that application of digital computer analysis to images obtained from rectilinear scanners and scintillation cameras would improve diagnostic quality. The primary difficulty was that the limited data acquisition matrices so degraded the spatial resolution of the imaging device that no amount of image processing could restore it to anything near the original analog image.

Image filtering consists of modifying the original image by logically 're-imaging' it with a mathematical imaging device in which spatial response can be controlled by the user. The difference between a logical and a real imaging device (for example, scintillation camera) is that the response function of the logical device can have negative values, whereas that of the real device can only be positive. The practical result of this difference is that the logical imaging device used for image filtering can be specified to improve spatial resolution and make edge and count density transitions more obvious. Alternatively, the response can be specified so that filtering smooths the image and reduces the noise so that small differences in count densities can be more readily perceived by the observer.

Image filtering is performed by a mathematical operation called convolution, which is simply the successive replacement of each point in the original image by a new value produced by a weighed combination of the original point and its surrounding neighbour points. Using larger filter functions, such as 5×5 or 9×9, will cause

larger portions of the original image to have an effect on the value of the new image points.

2.2 Cardiology

Cardiac studies represent another principal application of computers in modern nuclear medicine. A number of factors evaluated through the use of radionuclide imaging procedures include right and left ventricular ejection fraction, cardiac output, wall motion, cardiac size, intraventricular shunt quantitation and myocardial ischemia.

The gated cardiac study, perhaps the most visually dramatic, is used for ejection fraction calculation, cardiac output, and evaluation of wall motion abnormalities. In the gated study, data collection is synchronized with the cardiac cycle through the electrocardiogram signal. The data collection is set up in such a way that a series of images representing short segments of the cardiac cycle is collected in the memory of the computer. At the occurrence of the QRS complex (end-diastole), the computer begins data collection in image one. At relatively short intervals (20-40 ms), the data collection is moved to the next image in the series until the number of desired segments are collected or the cardiac cycle restarts with another QRS complex. The data is collected over a large number of cardiac cycles so that the images represent an average cycle rather than any one single cardiac cycle.

The first analysis step for this study is to replay the sequence of images in a movie-type display in which the viewer is given the impression of a beating heart and the clinician is allowed to evaluate, subjectively, the cardiac size and wall motion uniformity. The opportunity to obtain an overall subjective impression of cardiac function is also provided. This study is a perfect example of the ability to obtain a relatively large amount of clinical information from a computer-augmented study without a great deal of sophisticated quantitative analysis software that depends on a programmer's ability to anticipate every eventuality in the clinical study.

Ventricular ejection fraction is obtained from gated study data by calculating the relative differences between the volume of end-diastolic and end-systolic ventricular images. Simple ejection fraction, acquired by ventricle definition and background by hand-drawn regions of interest (ROIs), is usually adequate and often as reproducible as any automatic analysis method. For cases, however, in which a more objective analysis or a background-corrected ejection curve is desired, it is necessary to have automatic and objective methods that define the ventricular outline. A number of programs have been written for this purpose. The basic difference between most versions of these programs is the method used for definition of the ventricle edge.

Automatic edge detection techniques have been receiving a great deal of attention in computer analysis of clinical images. In an attempt to provide objective and automatic methods for organ definition in a number of different applications (i.e., renal size and function analysis), the detection of edges in a clinical image is based on a number of criteria, the most obvious of which is count level or threshold. Using this criterion, pixels above the cutoff threshold intensity are considered to be inside the organ of interest and those below to be outside. This is a good edge detection method for high contrast images, such as in renal studies, but has problems in studies

that have shifting background intensity. In this case, varying background intensities may confuse the decision because of one side of the organ of interest where background itself may be above the threshold, a situation that is often found in cardiac imaging. When this situation occurs, a second criterion can be used in making decisions about location of the edge. The derivative of signal intensity can be used to ask questions about the rate of change of levels in the image. The inflection point in which the curve changes from concave upward to concave downward is also the place in which the derivative is a local maximum or minimum. By taking the derivative, which is the difference between successive pixels, one can locate the inflection points representing the organ edge. Determination of left or right ventricle edges is usually a combination of these two techniques in that the threshold is used on the outer border while the inflection point is used in the septal region between the ventricles.

Intraventricular shunt determination is accomplished by an analysis of the time rate of activity change as a bolus of activity flows through the cardiac chambers. Once the bolus of activity passes through the right ventricle, it goes through the lungs and then back through the left heart. If there is an intraventricular shunt, blood flows from the left to the right ventricle without going through systemic circulation. Existence of a shunt is indicated by the reappearance of activity in the curve obtained from an ROI placed over the lung before enough time has elapsed for the bolus to have passed completely through systemic circulation. In case of a normal heart, the downward side of the bolus curve will not be distorted by the second peak. The analysis of the lung curve for calculation of shunt magnitude consists of fitting a theoretical curve to the primary peak and subtracting the fitted curve from the original lung curve. A second theoretical curve is then applied to this difference curve. The magnitude of the shunt is specified by the ratio of pulmonary blood flow to systemic blood flow (Q_p/Q_s). Pulmonary blood flow is determined from the area under the primary peak of the lung bolus curve and systemic blood flow is found by subtracting the area under the second peak from that under the first. Thus the ratio

$$Q_p/Q_s \quad \frac{\text{Counts under peak one}}{\qquad\qquad}$$

Although this shunt determination is conceptually a simple analysis process, it is usually complicated by practical considerations such as the inability to obtain good lung bolus curve because of patient positioning, and statistical noise in the curve which interferes with the fitting of the second peak.

The computer can be used to obtain quantitative evaluation of myocardial distribution of 201-*Tl* in the evaluation of myocardial ischemia. In this study, the patient is first imaged within minutes of the injection of 201-*Tl* tracer, often during an exercise test. A baseline image for determination of initial perfusion of the myocardium is provided. After a delay period to allow redistribution of the thallium the patient is imaged one or more times. Without computer support the subjective analysis of this study consists of a visual estimation of the amount of redistribution of tracer in the scintigrams. The computer, however, not only can provide a quantitative comparison of the images, but can also realign images that may vary in exact orientation because of difficulties in patient positioning. Quantitative analysis may include the

radial location of defects in the myocardial image as well as precise numerical estimates of the amount of change in the defects.

2.3 Flow Curve Deconvolution

The use of radioactive indicators as monitors of organ function extends back to the very beginning of nuclear medicine. The early measurements of renal and cardiac functions were made using stationary probes positioned at the appropriate place on the patient's body. Probe data provided a great deal of clinical information that could otherwise not be obtained. However, uncertainties in the location of the internal organs and the exact field of view of the probes introduced significant questions about the reproducibility and accuracy of flow measurements. The introduction of the computer interfaced scintillation camera provided a powerful clinical tool for more accurate analysis of organ function and flow studies. The data collection used for functional evaluation is a dynamic study with the frame rate specified so that desired details of the organ function could be detected. For renal evaluation, 1 frame every 30 seconds is common, whereas, as many as 10 frames every second are sometimes used for cardiac shunt studies.

Following the data collection, the region representing the organ to be evaluated is outlined with an ROI and the flow curve reconstructed. The resulting flow curve is essentially the same as that which could be collected from a probe. This curve is not directly representative of the organ function because the bolus used to produce the curve is usually introduced into the patient at some distance from the organ of interest. If a very narrow bolus is introduced into any organ system, the activity in the organ output will have some finite width and shape that is determined by the function of the organ. If the input bolus is already distorted by passage through other organ systems, the shape of the output curve from the organ of interest will have a combination of input bolus distortion and organ function. For a renal study, the injection of the bolus in the patient's antecubital vein means that the activity must move through a significant portion of the blood stream, the heart, and a number of other blood vessels before reaching the kidney. The practical effect of this procedure is to expand the bolus from the short, sharp spike of activity at the injection site to a fairly broad, smoothed-out bolus at the entrance to the kidney. In order to properly evaluate kidney function, it is necessary to compensate for this extended bolus by a mathematical process called deconvolution, similar in concept to the mathematics of image filtering. It is equivalent to filtering the organ output curve with a filter that has been designed so that its application to the input bolus would result in the bolus of zero width.

The process of performing such a deconvolution in clinical patient studies is not as easy as this short discussion would have, because a number of factors complicate the problem. Statistical uncertainty in both the bolus curve and the organ output curve causes mathematical failure in some cases. Furthermore, the determination of the input bolus shape is sensitive to the size of the ROI and its relative location to anatomical structures that may also contain radioactive tracer.

2.4 Single Photon Emission Computed Tomography

The true power of the computer is probably best exemplified by the tomographic capabilities provided by single photon emission computed tomography (SPECT) imaging systems. The ability to provide high contrast images of internal organs without the interference of overlying activity distributions gives the clinician the capability of evaluating low contrast structures that might otherwise be obscured.

The data collection process of SPECT imaging generally consists of the collection of a number of routine planar images (32-128) taken at sequential angular positions about the patient. These planar images are then processed to produce tomographic views. Although a number of reconstruction methods have been used in the past, the one most commonly used today is filtered backprojection. If the activity is imaged with a scintillation camera at different orientations, then profile lines drawn through each of the camera images at the source level will have shapes where a deflection from the baseline indicates an increased number of detected counts.

The simplest reconstruction procedure would be to simply spread (backproject) the counts from each of these projection lines back along the path from which they could have come. Because there is no *a priori* basis for believing that one location along this backprojection line is more likely to be the source of the counts than any other, this uniform redistribution of counts is as reasonable as any other distribution. This simple backprojection only results in a very rough approximation of the original circular source with a relatively high background 'star' effect. However, if the original raw data collected in the imaging process is first filtered so that each of the projections has a small negative component on either side of the peak, then the backprojection will produce an image free from spike artefacts. In this case the negative components of one projection tend to partially cancel the positive component from other projections, resulting in a much more faithful reproduction to the circular source and a reduction in the significance of the background star artefacts. Although clinical tomography uses a larger number of planar views, the concept of tomography image reconstruction is identical to this primitive example.

The choice of filter to be used for modifying the data prior to backprojection is dependent on a number of factors. The filter may be adjusted by the operator to enhance either spatial resolution, with associated increase in counting noise, or enhance low contrast lesion detection with a corresponding reduction in image noise.

2.5 Array Processors

With the advent of ECT (also called SPECT) and digital image filtering, nuclear medicine computer systems are now being called upon to support increasing data processing loads, in terms of both volume and complexity. An alternative to purchasing a new computer to cope up with this expanded work load is to add an array processor, since these can be obtained for a fraction of the cost of a new computer system. Array processors are general purpose add-on devices designed to perform high-speed mathematical computations such as fast Fourier transform (FFT) on arrays of data. When incorporated into a computer system they can reduce program execution time considerably if repetitive operations are to be performed on several collections of

data. They are, therefore, well suited for performing operations on whole nuclear medicine images as is the case with ECT and SPECT imaging systems.

By far the most important application of an array processor in nuclear medicine will be in reconstructing images for ECT. Use of an array processor can reduce reconstruction time for a 64 × 64 matrix and 64 angles from 15 to 30 s/section to about 1 to 2 s/section. An exciting application of array processors is to decrease the execution time of some of the more complex tomographic reconstructive algorithms, such as iterative reconstruction, to the point where they can be applied in clinically acceptable time. In addition to speeding the reconstruction of tomographic images, an array processor can be used to filter two-dimensionally the acquired data before reconstruction, and filter the images optimally after reconstruction. Thus, for example, marked suppression of noise can be obtained with two-dimensional FFT preconstruction filtering of the acquisition data.

In cardiac nuclear medicine imaging, a fast array processor has played an important role in obtaining functional phase image of the heart. Typically, a fast array processor has been used to do spatial and temporal Fourier filtering of a 32-frame, 64 × 64 pixel, gated blood-pool study and to form an extensive functional image set and histograms, all in less than 1 min. The functional image set includes the phase image, three images derived from the first and second derivatives of the time activity curves of individual pixels in the ventricular ROI, and a 15-frame series of dynamic ejection fraction images. The three images based upon the derivatives of the time activity curves allow separation of systolic and diastolic function, since they include information from more than the first harmonic.

Besides the formation of functional images, an array processor can be used for the digital filteration, both spatial and temporal, of dynamic images. With flow studies, this can provide a significant improvement in image quality.

Optimal digital filtering of static images can also be done rapidly using an array processor. Such techniques as Weiner filtering or stationary and non-stationary count dependent filtering in the frequency domain become feasible in clinically reasonable time period when an array processor is used.

The high-speed of digital filtering using an array processor makes it possible to filter an image under real-time control of the operator. An example of this is to vary the cutoff frequency of a Butter worth low-pass filter with a joystick, while watching the video display to select the cutoff frequency most appropriate for that image. As a typical case, one can implement this on a nuclear medicine computer system by using assembly language coding on an array processor of a two-dimensional FFT for real-time data, which transforms a 128 × 128 pixel image in less than 0.2 s. This allows for a new 128 × 128 pixel image to be written back to the video display approximately two times a second.

Filtering in the spatial domain is also much faster with an array processor. For example, it requires 0.4 s to pass a 64 × 64 pixel image to the array processor and perform a two-dimensional convolution of the image with a 13 × 13 pixel mask, against 8 s to carry out the same filtering on the host with an assembly language implementation of a new algorithm that minimizes the number of multiplications.

An innumerable applications of array processors exist in nuclear medicine imaging. The applications are limited only by the imagination of the programmer. The speed and processing power of array processors niakes them attractive additions to computer systems to enhance throughput and allow the use of processing algorithms that would otherwise be too time-consuming for a clinically dedicated nuclear medicine computer system.

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