Mr. 2204
CONF-720430

MASTER

**PROCEEDINGS OF SECOND SYMPOSIUM ON SHARING OF
COMPUTER PROGRAMS AND TECHNOLOGY IN NUCLEAR MEDICINE
OAK RIDGE, TENNESSEE, APRIL 21-22, 1972**

1972

**OAK RIDGE NATIONAL LABORATORY
OAK RIDGE, TENNESSEE**

DISTRIBUTION OF THIS DOCUMENT IS UNLIMITED

NOTICE

This report was prepared as an account of work sponsored by the United States Government. Neither the United States nor the United States Atomic Energy Commission, nor any of their employees, nor any of their contractors, subcontractors, or their employees, makes any warranty, express or implied, or assumes any legal liability or responsibility for the accuracy, completeness or usefulness of any information, apparatus, product or process disclosed, or represents that its use would not infringe privately owned rights.

This report has been reproduced directly from the best available copy.

Available from the National Technical Information Service, U. S. Department of Commerce, Springfield, Virginia 22151.

Price: Paper Copy \$6.00
Microfiche \$0.95.

PROCEEDING OF SECOND SYMPOSIUM
ON
SHARING OF COMPUTER PROGRAMS AND TECHNOLOGY IN NUCLEAR MEDICINE

Oak Ridge, Tennessee

April 21-22, 1972

Papers Presented at the topical meeting
of the Society of Nuclear Medicine, Inc.

Compiled by

Jane Gurney and Betty F. Maskewitz, ORNL

Symposium Organizing Committee

Lawrence K. Akers, ORAU
A. Bertrand Brill, Vanderbilt University
Francis H. Clark, ORNL
Betty F. Maskewitz, RSIC-ORNL
Henry N. Wagner, Jr., SNM

Cosponsored by

Oak Ridge Associated Universities
Oak Ridge National Laboratory*
and
Vanderbilt University

*Operated by
Union Carbide Corporation
for the U. S. Atomic Energy Commission

NOTICE

This report was prepared as an account of work sponsored by the United States Government. Neither the United States nor the United States Atomic Energy Commission, nor any of their employees, nor any of their contractors, subcontractors, or their employees, makes any warranty, express or implied, or assumes any legal liability or responsibility for the accuracy, completeness or usefulness of any information, apparatus, product or process disclosed, or represents that its use would not infringe privately owned rights.

PREFACE

Following the April 1971 symposium, the Computer Committee of the Society of Nuclear Medicine met to investigate ways by which the nuclear medicine community's anticipated increasing employment of computer technology may best be served. Among the proposals was a second symposium to focus on specific problem areas and attempt to determine goals and how best to achieve them.

The current symposium entitled "Second Symposium on Sharing of Computer Programs and Technology in Nuclear Medicine" was held on April 21-22, 1972 in Oak Ridge, Tennessee, with the Oak Ridge Associated Universities (ORAU), the Oak Ridge National Laboratory (ORNL) and ORTEC, Inc. as local hosts. Joining as sponsors were Vanderbilt University and the Society of Nuclear Medicine (SNM). Participating in the meeting were 108 attendees representing 57 installations, 3 located in Europe.

The symposium papers are being published as the result of a continuing policy of each of the above sponsors to make available information which is likely to advance the state-of-the-art in an important technical field. Special appreciation is expressed to the individual authors and their employers who devoted much time and effort toward making the symposium a success and this publication possible; to Edward D. Aebischer and his staff at ORAU for preparing and printing the programs; to N. T. Bray and his staff at ORNL for preparing the camera-ready copy and performing other necessary publication functions; and to the USAEC Technical Information Center, who agreed to publish and distribute these papers.

Deserving of special recognition are the members of the local arrangements committees whose efforts contributed much to the success of the meeting. ORAU staff member Bernice Corn and ORNL staff members Henrietta Hendrickson and Mildred W. Landay efficiently handled clerical and other necessary functions.

We are especially grateful to Mrs. Edward J. Spitzer of Oak Ridge for hospitality extended to the wives of participants.

It is proper to give credit to the members of the SNM Computer Committee for the conception of this second symposium and for assisting in setting up the program. The present members of the Committee are:

<u>Name</u>	<u>Institution</u>
I. R. Bell	The Molecular Anatomy (MAN) Program, ORNL
A. B. Brill, Chairman	Vanderbilt University School of Medicine
Don Brown	University of Colorado Medical Center
Valerie Brookeman	University of Florida, Dept. of Radiology
Don Chaapel	IBM, Rochester, New York
F. H. Clark	Instrumentation & Controls Division, ORNL
Rex Hill	Washington University Biomed. Computer Lab.
Lester Levy	Long Island Jewish Hospital
Betty F. Maskewitz	Radiation Shielding Information Center, ORNL
Charles Metz	Argonne Cancer Research Hospital
Henry Wagner	The Johns Hopkins Medical Institutions

The interest shown by the participants in the mission of the symposium to advance computer technology in nuclear medicine caused the formation of two subcommittees to function as working groups in special areas. The Subcommittee to devise standards for hardware interface, chaired by F. H. Clark, consists of the following members:

<u>Name</u>	<u>Institution</u>
Ross Burrus	TENNELEC, Oak Ridge, Tennessee
R. Esteverena	ORTEC, Oak Ridge, Tennessee
Russ Heath	USAEC Idaho Operations Office, Idaho Falls
Rex Hill	Washington University, St. Louis, Missouri
Judd Parker	Vanderbilt University, Nashville, Tennessee
Arthur F. Shufelt	Medical Data Systems, Detroit, Michigan
Charles Williams	ORTEC, Oak Ridge, Tennessee
James Woody	Instrumentation & Controls Division, ORNL

The Documentation and Programming Standards Subcommittee is chaired by W. J. McClain, ORNL, Oak Ridge, with current members:

<u>Name</u>	<u>Institution</u>
Joan Correll	Massachusetts General Hospital, Boston
W. M. Flowers	University of Mississippi at Jackson
Betty F. Maskewitz	Radiation Shielding Information Center, ORNL
Earl McDow	ORAU Computer Center
Ron Price	Vanderbilt University

CONTENTS

	<u>Page</u>
Preface	iii
Analysis and Correction of Spatial Distortions in Gamma Camera Images Valerie A. Brookeman, Steve S. Spector, Chester D. Klystra, and Nils J. Diaz	1
Interactive Image Manipulation System and Image Manipulative Extension to Higher Level Language for Use by Non-Computer Oriented Personnel Jon Erickson and Shelbourne Wilson	15
Recent Advances in Digital Processing Static and Dynamic Scintigraphic Data Dennis L. Kirch and Donald W. Brown	27
Collection, Processing, and Display of Dynamic Functions Nathaniel M. Alpert, Joan E. Correll, Linda A. Deveau, and Marc-David Cohen	55
Interactive Pattern Analysis and Classification Systems Laveen Kanal	63
Instrumentation for Pattern Recognition and Associated Software Robert S. Ledley and Louis S. Rotolo	69
The CAMAC Standard and Its Use in Computerized Measuring Systems K. Cúljat, R. Esteverena, T. Douglass	93
A Scanning System with Two Data Channels, Data Reduction and Display P. R. Bell, W. J. McClain, D. A. Ross, J. K. East, and E. C. Moore	109
The Gamma-11 System, a New Tool in the Nuclear Medicine Department John Silber	125
Basic Considerations in Computer Use for Dynamic Quantitative Radionuclide Studies Robert H. Jones, Bryce B. Bates, Jack K. Goodrich, C. Craig Harris	133
Anatomic Resolution in Dynamic Radionuclide Studies by Computer Identification of Radioactivity Fluctuation with Time Robert H. Jones, Ronnie B. Klaphaak, David C. Sabiston, Jr.	151
Clinical Applications of Computers - Nuclear Angiocardiology D. D. Watson, J. P. Nelson, S. Gottlieb	169
Investigation of Cardiac Dynamics Using a Scintillation Camera and Dedicated Computer Richard A. Ponto, Marvin E. Goldberg, Merle K. Loken	185
Increasing the Information Content in a CRT Display for a Nuclear Medicine Computer System Philip A. Miller, Donald R. Strange, Arthur F. Shufelt	199
Matching Gray Scale Display Programs to Various Film Types James C. Carlson	207

	<u>Page</u>
Computer Controlled, Intensity Modulated CRT Display of Isotope Scan Data Robert O. Smith	215
Sharing of Algorithms for Scan Processing in Nuclear Medicine A. E. Todd-Pokropek	223
Current Studies with the IDA (Image Display and Analysis) System Henry N. Wagner, Jr. and T. K. Natarajan	233
A General Logical Structure for Quantitative Analysis of Radiocardiographic Data David M. Shames	245
Localization of Heart Chambers by Temporal and Spatial Analysis for Regional Myocardial Blood Flow Measurements William J. MacIntyre, Robert E. Botti, Yasushi Ishii, and Thomas S. Houser	253
Computer Medicine, Phase II: From Innovators to Implementers Keith W. Sehnert	271
Programming Systems for Optimal Radiotherapy: Progress and Prospects Judith M. S. Prewitt	281
Radiation Transport Codes for Potential Applications Related to Radio- biology and Radiotherapy Using Protons, Neutrons, and Negatively Charged Pions T. W. Armstrong	303
Rapid Manipulation and Display of Counts vs. Time Curves by Hardware and Software Thomas F. Budinger	305
Some Guidelines for Documentation Practices as an Aid in the Sharing of Computer Programs and Technology Betty F. Maskewitz and W. J. McClain	325
Code Conversion by Computer A.E. McDow	335
Plan for Nuclear Medicine Communications Network Francis K. Herbig, Robert M. Donati	343
Data Acquisition and Display Techniques Samuel Wilensky	351
The Use of Video Systems for Acquisition and Display of Static and Dynamic Function Images A. B. Brill, P. Richardson, J. Erickson, J. Patton, and J. Parker	359

	<u>Page</u>
OTHER PAPERS SUBMITTED FOR PRESENTATION	
Gamma-Camera-to-Computer Interfacing -- The Easy Way T. R. Barclay, A. C. Morris, Jr., A. E. McDow, Jr.	369
Fast Data Acquisition with a Camera-Computer-System F. Bitter, W. E. Adam	377
Derivation and Computer Display of Regional Function Scans of the Lung with ¹³³ Xenon William J. MacIntyre and Scott R. Inkley	385
A Proposed Method for Quantitative Brain Blood Flow Transverse-Section Scanning J. A. Patton, A. B. Brill, and P. H. King	399
The Heidelberg Scintigraphic Image Processing System P. Pistor, P. Georgi, and G. Walch	411
Preliminary Results on the Use of the Hadamard Transform G. Berche, A. E. Todd-Pokropek, and R. Di Paola	429
Serial Scan Data Analysis by "Mini-Computer" Robert O. Smith, Margaret S. Boyd, Patrick H. Lehan, Harper K. Hellems	439

ANALYSIS AND CORRECTION OF SPATIAL DISTORTIONS
IN GAMMA CAMERA IMAGES

Valerie A. Brookeman, Steve S. Spector
Chester D. Klystra and Nils J. Diaz

ABSTRACT

Spatial distortions in radionuclide images from a gamma camera interfaced to a PDP-8/I computer were investigated by analyzing the system response to an orthogonal ^{99m}Tc grid source using an IBM-1800 computer. Each intersection point of two line sources was determined and a non-linear spatial distribution observed. Weighted spatial errors were calculated for all 2000 matrix locations in the camera's field of view and ranged from -0.9 to +1.1 location (6.5 mm). A program, developed for the IBM-1800 to correct any image recorded with the camera/PDP-8/I system of spatial distortions, has been applied to various clinical images.

University of Florida and Veterans Administration
Hospital, Gainesville, Florida

April, 1972

The existence of spatial distortions in gamma camera-produced radionuclide images has been recognized and observed by several workers,¹⁻⁷ but corrections for spatial distortions in clinical images are not currently made. We have investigated the spatial distortions in radionuclide images caused by the gamma camera, determined the effect of flood corrections for non-uniform sensitivity response of the camera on spatial linearity, and developed a method for spatial error correction of any camera image by digital computation.⁸

EXPERIMENTAL METHOD

Our Pho-Gamma scintillation camera is interfaced to a PDP-8/I digital computer⁹ with 8K of core memory and each image is represented as a 50 X 50 data matrix. Peripheral equipment consists of a Teletype printer-keyboard and paper-tape read/punch for input of commands and programmes, and output of data; an oscilloscope for data display; a high speed line printer for fast data output; and 2 DEC tape units for programme and data storage.

An orthogonal grid of ^{99m}Tc line sources, 3 cm apart, (Figure 1) is placed at the face of the 4,000-hole technetium collimator, with the line sources parallel to the crystal axes, and 3 million counts collected by the camera-PDP-8/I system. Forty-five intersections of the perpendicular line sources are in the camera field of view as seen in Figure 2.

The digitized grid image may be corrected for non-uniform sensitivity response of the system by applying correction factors from a ^{99m}Tc crystal flood¹⁰ and is then punched on binary perforated tape and transferred to an IBM-1810 disc. The spatial error analysis and correction routines are currently performed on an IBM-1800 computing system with 12K of core memory, due to the present limited data handling facilities of our PDP-8/I.

The following data processing I/O devices were used with the IBM-1800 computer: (1) IBM-1816 printer-keyboard, (2) IBM-1442 card read/punch, (3) IBM-1810 disc storage, (4) IBM-1627 plotter, and (5) IBM-1054/1055 paper tape read/punch. All programmes for spatial error analysis and correction were written in Fortran IV for the IBM-1800.

SPATIAL ERROR ANALYSIS

A flow diagram of spatial error analysis is shown in Figure 3. The 45 intersection points of the line sources are identified by the computer and 3 X 3 matrices representing each intersection are stored. Each 3 X 3 array is fitted by least squares to a two dimensional asymmetric model and the position of the maximum value determined. This is the exact position of each intersection, as seen by the camera. The mean spacing D between any two adjacent maxima is computed and an ideal theoretical grid constructed with constant line spacing D. The deviations in X and Y directions of each intersection point as seen by the camera from the theoretical grid are determined and Figure 4 shows an example of a grid source image, with and without flood corrections. The X and Y deviations of the determined intersection points from the ideal grid are summarized in Table 1, with and without flood correction factors applied to the grid image.

The X and Y deviations from the theoretical grid are computed for all locations in the data matrix as follows. The X deviations of 8 intersection points, in a 4 X 4 array, are fitted to a two dimensional model and an X deviation determined for every matrix location in the central square of the 4 X 4 array. All intersection points are considered in turn, grouped in 4 X 4 arrays, until an X deviation has been determined for every matrix location over the camera face. Similarly, Y deviations are determined for all matrix locations. The two resulting error matrices of X and Y non-linearities are stored on disc for future use in correcting the spatial distortions introduced by the gamma camera/PDP-8/I system to any image.

The deviations from linearity of all data matrix points over the camera face are shown in Figure 5, and the directions of the X and Y deviations in Figure 6. (+) and (-) signs indicate deviations in the positive and negative

X or Y directions. Figure 7 shows the actual magnitudes of the X and Y deviations of all data matrix points expressed as the percentage of matrix points with a certain spatial deviation in units of matrix channels (1 matrix channel is about 5.6 mm). The generation of two spatial error matrices, for both X and Y directions, currently takes 5-8 minutes with the IBM-1800 computer, excluding any displays or plots of the magnitudes and bias of the spatial errors. The time for actual programme execution is 2-3 minutes, the additional time being due to pauses for display options.

SPATIAL ERROR CORRECTION

Any image recorded by the camera/PDP-8/I system may be transferred to disc and corrected of spatial errors by either of two methods A or B using appropriate spatial error matrices. The resulting corrected image is stored on disc and may be displayed in several formats. Method A shifts each data point in both X and Y directions to its correct position, leaving the count content unchanged. This results in non-uniform intervals between adjacent data points, and for actual display of an image is impractical.

A flow diagram of spatial error correction by method B is shown in Figure 8. Each data point is left in its original uniform-interval position, and a corrected count calculated for each point, taking into account the spatial deviation of each point from its true position. Each data matrix point is considered in turn as the central element of a 3 X 3 array of nine locations, in their true spatially correct positions. Each 3 X 3 array is fitted to a two dimensional asymmetric model and a count content for the central element, in its original position as seen by the camera, is computed. This is done for all matrix locations over the camera's field of view and currently takes about 15 minutes per image, excluding any plots or displays.

Figures 9 and 10 show the results of spatial error correction by both methods A and B for two clinical images. Both Figures 9 and 10 are cross-sections through right lateral lungs after injection of 2 mCi of ^{99m}Tc -iron hydroxide aggregates. Since method B employs a least squares fitting technique to calculate a new count for each data point, the image undergoes two dimensional smoothing in addition to spatial error correction, as seen in Figure 9. In Figure 10, method B was employed for spatial error correction. The lower profile, corrected only for non-uniform sensitivity response, has a dip in the middle which could have been interpreted as an abnormal feature. This disappeared when spatial errors were also corrected, proving it to be an artefact.

SUMMARY AND CONCLUSIONS

Spatial distortions are introduced to radionuclide images by the gamma camera/PDP-8/I computer system. Application of flood correction factors for non-uniform sensitivity response reduces spatial non-linearities by about 30%. Any digitized image recorded with the camera-computer system may be corrected of these spatial distortions by analytical techniques, using software routines developed for an IBM-1800. Since the average spatial error of any data point is only 0.3 of a matrix channel, or about 1.8 mm, it is more important for laboratories with limited computer facilities to correct for the gamma camera's non-uniform sensitivity response, and for routine qualitative clinical interpretation of images, it would not appear necessary to correct camera images of spatial errors. However, even after flood correction, individual spatial errors can be large. One data point may have a positional error as large as 1.1 matrix channels, or 7 mm, and thus spatial errors are important and should be corrected if gamma camera data are to be quantitatively analyzed, such as in dynamic function studies.

TABLE 1

Deviations From Linearity of Grid Source Intersection Points

Sensitivity Corrections	X Deviation (mm)		Y Deviation (mm)	
	Range	Mean (SD)	Range	Mean (SD)
No	-8.0 - +6.7	2.4 (2.1)	-8.4 - +7.8	2.2 (1.9)
Yes	-6.5 - +5.7	1.8 (1.5)	-5.1 - +6.4	1.4 (1.1)

Figure 1. Orthogonal grid phantom (not to scale)

Figure 2. Line sources (3 cm apart) with respect to camera crystal

SPATIAL ERROR ANALYSIS

Figure 3. Flow diagram of spatial error analysis

SYSTEM RESPONSE TO GRID SOURCE

Figure 4. Line sources as seen by the system

SYSTEM DATA MATRIX

Figure 5. Data matrix as seen by the system

BIAS OF DEVIATIONS FROM LINEARITY

WITHOUT SENSITIVITY CORRECTIONS

WITH SENSITIVITY CORRECTIONS

Figure 6. Bias of deviations from linearity in the X and Y directions

—•— WITH SENSITIVITY CORRECTIONS
- - - WITHOUT SENSITIVITY CORRECTIONS

Figure 7. Distribution of spatial error in the data matrix

SPATIAL ERROR CORRECTION

Figure 8. Flow diagram of spatial error correction

Figure 9. Cross section through right lateral lung before and after spatial error correction by methods A and B

CROSS SECTION OF RIGHT LATERAL LUNG STUDY

Figure 10. Cross section through right lateral lung before and after spatial error correction by method B

REFERENCES

1. J. R. Mallard and M. J. Myers, "The performance of a gamma camera for the visualization of radioactive isotopes in vivo," Phys. Med. Biol., 8, 165-182 (1963).
2. T. D. Craddock, S. O. Fedoruk and W. B. Reid, "A new method of assessing the performance of scintillation cameras and scanners," Phys. Med. Biol., 11, 423-435 (1966).
3. M. J. Myers, P. J. Kenny and J. S. Laughlin, "Quantitative analysis of data from scintillation cameras," Nucleonics, 24, 58-61 (1966).
4. H. O. Anger, in Instrumentation in Nuclear Medicine, Vol. 1, G. J. Hine, Editor, p. 511 (Academic Press, New York, 1967).
5. E. Jahns and G. J. Hine, "A line-source phantom for testing the performance of scintillation cameras," J. Nucl. Med., 8, 829-836 (1967).
6. J. B. Svedberg, "Image quality of a gamma camera system," Phys. Med. Biol., 13, 597-610 (1968).
7. B. R. Westerman and H. I. Glass, "Physical Specifications of a gamma camera," J. Nucl. Med., 9, 24-30 (1968).
8. S. S. Spector, V. A. Brookeman, C. D. Kylstra and N. J. Diaz, "Analysis and correction of spatial distortions produced by the gamma camera," to be published in J. Nucl. Med.
9. F. P. Bruno, V. A. Brookeman and C. M. Williams, "A digital computer data acquisition, display and analysis system for the gamma camera," Radiology, 96, 658-661 (1970).
10. L. M. Morrison, F. P. Bruno and W. Mauderli, "Sources of gamma-camera image inequalities," J. Nucl. Med., 12, 785-791 (1971).

ABSTRACT

Interactive Image Manipulative System and Image Manipulative Extension to Higher Level Languages for Use by Non-Computer Oriented Personnel

Jon Erickson and Shelbourne Wilson

If image manipulative systems are to become viable in a purely clinical environment their capabilities must be available to the non-computer oriented personnel with a minimum of training. An interactive image manipulative software system for the PDP-9 overlay monitor is described. It provides access to image manipulations with default or user supplied parameter values. Operations may be chained to produce stored programs for later execution. An extremely flexible command format is used with error message and prompting statements for entry of parameter values. Examples of the use of the system are included. Also discussed is a proposed extension to high level languages for image manipulation. Various possible modes of implementation are discussed with emphasis on an interpreter mode for interactive languages such as BASIC and a converter mode for languages such as FORTRAN. This extension will allow generation of image manipulation programs while treating image arrays as single scalars.

Vanderbilt University
Nashville, Tennessee

Supported in Part by AEC Contract AT-(40-1)-2401

April 1972

**Interactive Image Manipulation System and Image
Manipulative Extension to Higher Level Language for
Use by Non-Computer Oriented Personnel**

Jon Erickson and Shelburne Wilson

Two desirable attributes which should be a part of the computational capability of any nuclear medicine facility which attempts to perform a significant amount of experimental manipulation and processing of radio-isotope images are: first, the ability to have easy access to and use of the large number of image manipulation programs that inevitably evolve in such a system, and secondly, the ability to rapidly produce specialized image analysis programs for specific analysis problems encountered in the course of a single research project. Ideally both of these capabilities should be available to the non-computer oriented personnel, such as the rotating medical staff, as well as the technical personnel such as the engineers and physicists. This paper describes an interactive manipulation system which provides the first of these capabilities and an extension to higher level languages which provides the second.

The interactive manipulation system will be considered first. The characteristics which we consider to be desirable in an interactive system which is to be used profitably by the uninitiated as well as the more sophisticated users are the following:

- (1) The system should be able to provide the majority of the information required to operate it.
- (2) The user should be able to enter a reasonable command without having to refer to a lookup list for the proper function code.
- (3) The system should respond with understandable error messages.
- (4) Default values for data dependent variables should be available to the user.
- (5) The experienced user should not be continually subjected to requests for information when he can provide it without prompting.
- (6) The system should be easily modified to allow for additions and alterations as new programs are developed.
- (7) The user should be able to chain together several of the manipulation routines to produce a "stored program" which can be run repeatedly without having to reenter each command each time.

We have implemented an interactive system which has all of these attributes and which for the purposes of this discussion will be described from an operational viewpoint.

The system operates under the overlay monitor of the PDP-9 (Digital Equipment Corp.) in which a main program remains resident at all times and the various functions are performed by subroutines which are brought into core from the disc as required. The only function of the main element in this system is to provide a common path between the command decoder and the operational routines which are called by the user. It also provides a permanent entry point for teletype interrupts which are initiated by the user to recover from unterminating program operations.

The decoder accepts three sets of commands. The first set contains those which consist only of the desired function, with or without arguments. The second set contains those in which the function name is preceded by a single character command modifier. The third set of commands are those which control the stored program capability of the system.

Function names may consist of any number of characters between 1 and 6. The only restriction on the name is that it must be unique to a single function. Only those characters included in the recognition code are required, and any others that may follow it are ignored by the system. The effect of this will be seen in the example shown below.

Arguments or program parameters are required for many of the image manipulation routines. While it is true that for routine clinical applications many of these parameters such as weighting factors for a smoothing operation can be fixed, this is certainly not true for a research environment where the user may wish to supply his own values. We have, therefore, chosen to implement a method by which we may have the best of both situations. When a routine is added to this system, default values are assigned to any data dependent parameters it may use. The user of the system may also supply his own values at the time he requests the operation. To enter argument values the user simply types them following the function name. At present these are restricted to signed integer values. The argument values may be separated by a space, a comma or a tab.

Figure 1 shows an example of the various forms a request might take. All of these will request the same photograph generation program. The first two forms will use the default values for the arguments. The last two will use the user supplied values. If nothing follows the function name, the default values are used. If, however, the user enters any character after the name, the system assumes he wishes to use other than the default values. The command decoder knows how many arguments each routine requires, and if the user enters the proper number no further response is indicated and the operational routine is loaded and executed.

However, two things will cause a prompting operation to take place as illustrated in Figure 2. Here the user wanted to request the photograph generation program. In the first case the wrong number of arguments were entered, and in the second a non-numeric character was entered. This figure also illustrates what we consider to be understandable error messages and prompting statements.

The second set of commands are those in which the user precedes the function by a modifying character as shown in Figure 3. Though there obviously must be a user's manual available for any system of this sort, it is very desirable to provide the user with the ability to obtain short descriptive paragraphs about the individual routines simply by asking the system for it. The first modifier shown here allows the user to do this. The user is instructed to see the line printer where the information is normally presented. The second modifier provides the user with the ability to obtain a listing of the prompting statements followed by the default values of the respective parameters. This allows the user to determine if he wishes to use the default values or would rather provide his own. The third modifier allows the user to indicate his desire to supply arguments and requests prompting without having to force it by deliberately entering erroneous characters.

The fourth modifier provides the user with ability to reassign the default argument values. This is useful when an operation will be performed repeatedly with other than the original default values. These new values remain effective only for the current system operation or until reset and do not effect the permanent system.

The utility of this type of system is enhanced considerably by the ability to chain several operations together to produce a stored program which may be recalled and run at any time without having to reenter the command strings each time. This system allows the user to create stored programs by specifying a program name (Figure 4) followed by the commands he wishes to include. The standard decoder operations take place during the storage so that the stored program has no errors from the standpoint of argument count. The program is run as shown in Figure 4 by requesting program execution and specifying the number of times it is to run. Any stored program may call any other. The programs are stored in a disc file and so are permanently added to the system, but may be deleted at will.

At any time the user may request a listing of the operations available in the system. An example of the list is shown in Figure 5. The operations available will not be discussed but they include data bounding, flat field flood correction, scanner speed correction, 9 x 9 smoothing, photograph generation, line printer output, ability to access all the bulk storage devices such as DECtape and disk

Figure 6 shows a typical operation of the system. The annotation to the right of the commands are self explanatory. This single example shows the versatility of the system and illustrates the use of the simple command structure to provide relatively complex image manipulations. This manipulation system is currently operational in our laboratory and has gained wide acceptance among the personnel. It provides all of the current image manipulation routines which we have available for static image analysis. It is extremely easy to use and allows the performance of an unlimited combination of manipulations with no restrictions on the order in which they are performed or the parameter values employed. It does not, however, provide the ability to produce new computational operations which may be required for some experimental analyses.

In order to provide this capability and to extend the philosophy of this system, we are currently engaged in designing and implementing a programming system providing the image manipulative capabilities of this interactive system but within the context of pre-existing computer languages. The adoption of a standard computer language as a base facilitates use by individuals having previous experience with computer languages yet retains the full power of the base language. While adaptable to essentially any general purpose scientific language, FORTRAN and BASIC are currently being considered. In practice the programmer will write his code treating variables which he has declared to be images (Figure 7) very much as though they were simple scalars. In addition he will have available new commands specifically for image manipulation. In the example shown in the figure, the variables FRAME through FLAG are declared in the first statement to be images with standard or default dimension. The executable code illustrated is a simple routine which reads a dynamic study frame by frame, provides flat field flood and background correction for each image and produces a dynamic record of the activity within a predefined region of interest. In addition each block of five consecutive frames is summed for display on an output device. The coding is straightforward, is independent of subroutine calling sequences, and is devoid of consideration of core allocation requirements. The majority of new commands are either non-executable specification statements defining variable type, image size and similar qualities, or else they are input/output directives. Examples shown are the IMAGE declaration and the READ IMAGE and DISPLAY commands. Most image manipulation within the program is accomplished using what appear to be conventional arithmetic statements and function calls.

Several possible levels of implementation are available. Implementation as a command decoder with statement by statement entry and execution is only slightly easier to accomplish than a free standing language subset, and offers no significant advantage over the interactive system discussed earlier. A command decoder capability is, of course, implicit in any implementation using a language such as BASIC. Implementation at a stand alone level offering stored program capability and conditional transfer of control provides essentially an image manipulative language. If the programming is done with a consideration for future expansion, then only the essential features need be included initially, with additional commands being added as the need arises. Current plans call for a BASIC based stand alone system to be the first of these language extensions to be brought up. The FORTRAN oriented system under development incorporates a rather different translation approach than does the BASIC system and will represent an integrated language with the FORTRAN language being a subset of the language.

A variety of implementation modes are available, each with characteristic advantages and weaknesses. Implementation may be as an interpreter, with storage of only the source language statements and execution concurrent with decoding; as a compiler with production of an equivalent machine dependent object program for later execution; or as a converter which reduces the augmented language source program to an equivalent standard language source program. An interpreter is ideally suited for interactive operation and this approach is being used in implementing the BASIC oriented system. The FORTRAN system, on the other hand, is being designed around a converter which will detect non-standard elements in a source program and convert the statements containing them into source code acceptable to the standard FORTRAN compiler. In large part, this is done by generating subroutine calls as illustrated in Figure 8.

In the example shown, a raw image is multiplied element by element using a flat field flood correction; the background is subtracted; and the image rewritten into core. The converter is responsible for house-keeping tasks such as work area allocation. In addition it may be required to provide additional functions such as dynamic core allocation or a rollin/rollout feature for virtual memory capability. Use of a higher level language for converter output (as opposed to the machine-dependent output of more conventional compilers) allows substantial machine transparent operation and does not require modifications to a major compiler, thereby simplifying the initial programming task.

We feel that if computer acquisition systems are to become viable in a purely clinical environment some means must be available for the clinician who is simply interested in the results of the routine tests to perform such tests and interact with the system without having to be concerned about the sequence in which he pushes a half dozen buttons or finding a pushbutton or key labelled with the function he desires. We believe that our interactive software system is a step in this direction. Of course, the level to which this is implemented is highly dependent on the hardware configuration, but some form of it is possible on any system containing a programmable computer and is of a size to allow image manipulation.

We further feel that the language extensions discussed here will allow those who are more clinically oriented to produce rather complex image and data analysis routines with a somewhat reduced responsibility for the detailed code production involved in such a programming endeavor. These language extensions, of course, are machine independent to the same extent as their base languages are machine independent. The adoption of systems such as these will ultimately reduce the complexity of operating computation and analysis systems and will be important in future developments in this area.

ACKNOWLEDGEMENT:

This work was supported in part by AEC Contract AT-(40-1)-2401

COMMAND EXAMPLE

```
>PHO  
>PHOTO  
>PHOT 2,100  
>PHOTOGRAPH 2 100
```

Fig. 1. Examples of the various forms a request for the same photograph generation program might take. The first two request the default parameters while the user supplies his own in the last two.

```
>PHOTO 4,5,6  
WRONG ARG CNT!  
PHOTO PTS/DATA PT = 4  
NORMALIZATION VALUE = 100  
  
>PHO %  
NUMBER ERROR!  
PHOTO PTS/DATA PT = 4  
NORMALIZATION VALUE = 100
```

Fig. 2. Illustration of the use of error messages and prompting statements to aid the user in calling a program. User entered text is underlined

COMMAND MODIFIERS

<u>EXAMPLES</u>		<u>OPERATION</u>
I PHOTO	- - - - -	USER REQUESTS INFORMATION ABOUT THE ROUTINE "PHOTO"
R PHOTO	- - - - -	USER WISHES TO KNOW DEFAULT VALUES OF THE PARAMETERS
X PHOTO	- - - - -	USER WISHES TO SUPPLY HIS OWN PARAMETER VALUES
A PHOTO	- - - - -	USER WISHES TO REASSIGN THE DEFAULT PARAMETER VALUES

Fig. 3. Examples of the four single character command modifiers are shown.

STORED PROGRAMS

CREATION OF PROGRAMS

>S FLOOD	/store a program called "FLOOD"
>BOUND 3	/data bound image to 3 stand. dev.
>SMOOTH 1,1,1	/3X3 smooth with equal weights
>FLAT	/flat field flood correct
>END	/end of program
>S CURVE	/store program called "CURVE"
>GET MT 25,0	/get 1 frame of study 25 from tape
>E FLOOD 1	/execute "FLOOD" one time
>TOTAL 25,25,30,33	/sum counts in this rectangle
>END	/end of program

EXECUTION OF PROGRAMS

>E CURVE	/execute "CURVE" 60 time to extract /data from 60 frames of the study

DELETION OF PROGRAMS

>D CURVE	/delete "CURVE"
>D FLOOD	/delete "FLOOD"

Fig. 4. Examples of the creation use and deletion of a stored program to extract data from a rectangular area of a 60 frame dynamic study stored on magnetic tape. Note the ability of one program to call another stored program.

SMO(OTH)
 BOU(ND)
 PHO(TOGRAPH)
 CON(TOUR)
 BCD(BCD TO BINARY CONVERSION)
 DIM(ENSION)
 LIS(T)
 NOR(MALIZE)
 GET DT,DK,MT,409(6),FIL(E)
 PUT DT,DK,MT,409(6),LP,CRT
 EOF DT,DK,MT
 FRA(ME)
 SHI(FT)
 ABP(HANTOM)
 FBE(FOREGROUND/BACKGROUND ERASE)
 TOT(AL)
 MAR(K)
 SPL(IT)
 SAV(E)
 RES(TORE)
 ERA(SE)
 LAB(EL)
 VSS(VARIABLE SIDED SMOOTH)
 VAV(VARIABLE AVERAGING)
 SPE(ED CORRECTION)
 GM(GEOMETRIC MEAN)
 CPH(ANGER CAMERA SIMULATION PHOTOGRAPH)
 EXP(ANDED CONTOURS)
 FF(FLAT FIELD FLOOD FACTOR CALCULATION)
 FLA(TTEN)
 9X9(9X9 CONVOLUTION SMOOTHING)
 GMP(ISORT(AB)/(ISQRT(A)+ISQRT(B)))
 CE(CONTRAST ENHANCE)

Fig. 5. Example of the list obtained on the line printer in response to the "LIST" request. Optional text or explanatory information is included in parentheses.

```

>E SCAN3                /request execution of SCAN3 system

EXECUTE V4A              /overlay monitor loaded

DO YOU NEED INSTRUCTIONS?(Y/N)
N                        /user requires no instructions

>GET DT -12              /read data in reverse from block 12 of
                        /DECTape
>BCD                     /BCD to binary code conversion of image data

>BOUND 3                 /data bound image to 3 standard deviations

>FF                      /calculate uniformity factors from image data

>GET DT -8               /read data in reverse from block 8 of DECTape

>SAVE                    /save current image in a disc file

>FLATTEN                 /correct uniformity with factors calculated
                        /above
>R PHOTO                 /request default argument values for photo
PHOTO PTS/DATA PT = 4    /generation program
NORMALIZATION = 50
>PHOTO 2,30              /request photo generation program with user
OPEN SHUTTER - CARRIAGE RETURN WHEN READY /supplied values

>RESTORE                 /retrieve image from disc file

>PUT NT                  /attempt to store on magnetic tape
UNKN PGM
>PUT MT                  /correct command for storage as study 35
STUDY ID = 35            /in cumulative mode
dyn=0,CUM=1
MODE=1

>EOF MT                  /write end of file mark on tape
>ERASE                   /erase image in disc file

> C                      /return to keyboard monitor system

KM9=15 V5A              /keyboard monitor reloaded

```

Fig. 6. Example of the use of the interactive image manipulation system.

```
IMAGE FRAME, DSUM, BACK, NORM, FLAG
```

```
DO 6 J=1, MAXNO, 5
  DSUM = 0
  DO 7 K = 0, 4
    READ IMAGE(110) FRAME
    FRAME = FRAME * NORM - BACK
    DYNAM(J+K) = CURVE(FLAG, FRAME)
7  DSUM = DSUM + FRAME
6  DISPLAY(109) DSUM
```

Fig. 7. Example of the use of the IMAGE declaration statement to declare that the indicated variables are images. The program segment shows the use of these image variable as simple scalars to produce a curve from data obtained by the READ IMAGE and display every block of five frames.

```
FRAME = FRAME * NORM - BACK
```

```
CALL MULTIM(FRAME, NORM)
CALL SUBIM(IMAC1, BACK)
CALL MOVIM(FRAME, IMAC1)
```

Fig. 8. Example of the output of a converter which translated the command shown into the three standard FORTRAN subroutine calling statements.

RECENT ADVANCES IN DIGITAL PROCESSING STATIC AND DYNAMIC SCINTIGRAPHIC DATA¹

Dennis L. Kirch and Donald W. Brown

ABSTRACT

One of the major problems associated with processing static scintigraphic images as well as dynamic time-activity curves concerns the varied statistical and functional nature of the data. This paper presents an adaptive smoothing/enhancement technique for processing static images. An intercomparative technique for evaluating image enhancement filters is discussed and examples of clinical images are shown. An interactive display oriented technique for analyzing time-activity curves is also described and an example of its adaptability to processing cardiac flow studies is presented. The overall emphasis of the paper concerns the need for flexible techniques in processing static and dynamic scintigraphic data using digital computer methods.

U.S. Veterans Administration Hospital and the University of Colorado Medical Center

¹This work was supported by V. A. Research Grant 01/1103.1/69-01

RECENT ADVANCES IN DIGITAL PROCESSING STATIC AND DYNAMIC SCINTIGRAPHIC DATA

INTRODUCTION:

Work performed at the Denver V. A. Hospital has resulted in the development of some advanced digital computer techniques for analyzing and processing static and dynamic scintigraphic data. While the respective problems and methodologies associated with processing static images and dynamic time-activity curves seem to be unrelated, there are some shared concepts that have value in both areas of application. The most significant difficulty associated with processing static and dynamic data concerns the diverse statistical and functional nature of routine clinical studies. As a result, there is an ever present need to adjust the parameters and techniques used in digital processing in order to reconcile the analytic methods to the count density and frequency composition of the data. The approaches to operating on static and dynamic data which will be discussed here share a common aspect in that they provide an adaptive flexibility which, to some extent, makes them applicable to a larger class of data than previous methods.

In the area of static image processing, we have explored the potential benefit of smoothing and enhancement of static images by linear and non-linear digital filtering methods. This work has made it apparent that linear techniques rapidly approach a point of diminishing returns with respect to the resolution enhancement achievable without excessive creation of artifacts. As an outgrowth of this work with linear filters, we have developed a non-linear adaptive smoothing and enhancement technique which is able to extend the resolution enhancement capability achieved in our earlier work with linear techniques.⁽⁷⁾ One particular advantage of this adaptive non-linear technique is that it is self-optimizing in the sense that the proper degree of smoothing is applied to each particular image with respect to the statistical nature of the image. This feature eliminates the undesirable necessity of modifying the smoothing parameters of the linear technique to fit the statistical make-up of the data.

Another development has occurred in our work with digital processing and analysis of the time-activity curves extracted from dynamic scintigraphic data. The varied functional and statistical nature of these time-activity curves presents some similar problems to those encountered in processing static image data. In addition, it is often necessary to perform a variety of operations on one or a combination of these curves in order to compensate for such data contaminating factors as time variable background, bolus spread function, and limited temporal resolution. For the above mentioned reasons we have developed an interactive, display oriented curve analysis program which places the general mathematic functions necessary to perform these operations at the finger tips of the operator. This technique has been developed to the point where the analysis can be conducted on a "single-step" cut-and-try basis or it can be programmed into an extensive protocol capable of performing a complicated series of operations with or without human intervention.

DESCRIPTION OF DEVELOPMENTAL COMPUTER SYSTEM:

Although the subject being described here primarily concerns digital techniques which have general computer applicability in processing scintigraphic data, a parallel

effort conducted in our laboratory has resulted in the development of a computer system on which these techniques have been implemented and verified. This system has been configured to meet the most demanding clinical requirements of current Nuclear Medicine procedures in terms of the accumulation, storage, retrieval and display of scintigraphic data. However, it is primarily intended to serve as a flexible laboratory tool for developing new analytic and diagnostic techniques. The system consists of a PDP-12 computer with 8K words of core memory and an 800K word magnetic disk. The computer has been interfaced to a Pho/Gamma HP Anger camera with a Computer Corporation of America interface. Static imaging is performed using a 64 x 64 element image array and dynamic studies are performed with a 32 x 32 array at rates up to 10 frames per second and a duration of up to 500 frames per study. One of the essential features contributing to the flexibility of this system is the ability to access and operate on scintigraphic data with Fortran. Furthermore, machine language instructions can be interspersed with the Fortran statements in the same program to speed up the computation time where it is expedient to do so. The interactive time-activity curve analysis program referred to earlier is also easily programmable using a highly simplified command language which can be easily mastered by persons without computer experience.

INTERCOMPARISON OF LINEAR FILTERS

The first successful application of digital filtering to processing static scintigraphic images involved the well known concept of two-dimensional averaging or smoothing. In routine clinical images having count densities of less than 1000 counts/cm² the magnitudes of the statistical fluctuations are significantly reduced by such spatial weighted smoothing operations as have been proposed by several investigators.⁽¹⁻⁶⁾ Unfortunately, this early work made little use of such analytic determinations of filter performance as frequency domain "side-lobe level," equivalent noise bandwidth, Gibbs phenomenon, or signal-to-noise ratio. As a result, most of the scintigraphic image processing work conducted by various investigators in the past has not been examined comparatively due to lack of enough analytic common ground to serve as a basis for such an intercomparison.

More recently, workers in this area have begun to look at digital filtering as more than a means of limiting the statistical fluctuations present in digitized images. A number of linear digital filter functions have been proposed⁽⁷⁻¹²⁾ which combine the capability for smoothing with some degree of resolution enhancement or refocusing of the image data. The purpose of this is to provide some manner of compensation for the limited resolution of existing scintigraphic imaging systems. While such smoothing/enhancement filters can also be implemented by two-dimensional weighting matrices, the larger dimensions of the matrices involved make them less attractive than simple smoothing filters from a computational standpoint. The analytic derivations underlying most of the proposed smoothing/enhancement filters have been more extensive. However, intercomparison of these techniques is still a formidable undertaking.

While the full potential of digital processing of static images is as yet unresolved, some initial work has indicated that digital filtering is a highly useful clinical tool. The diagnostic benefit of digital filtering in terms of lesion detectability has been well demonstrated by the very elaborate and complete simulation experiment conducted by Maroglio, Knowles, Kohlenstein and Schulz.⁽¹⁶⁾ In their work at the Applied Physics Laboratory, 4,000 simulated organ scans containing various sized lesions were read by trained observers. The results of this experiment indicated that the increased probability of detection amounted to 10% for smoothing alone, while smoothing followed by collimator compensation using a matrix inversion technique provided an additional 15% improvement. Unfortunately, the amount of equipment and personnel required to conduct an end-to-end experiment of this extent are not available to many of the experimentors who would like to evaluate various aspects of digital filter development.

In retrospect, it can be said that little success has occurred in the development of a quantitative measure of digital filter performance in terms of increased lesion detectability or minimization of false detections. While several formal mathematic criterion have been proposed for measuring the detection capability of the overall scintigraphic imaging system, (13-15) the dependence of these criterion on scanning system parameters would require that they be completely re-defined to include filter function parameters. Also, the problem of modeling the performance of the human observer presents additional levels of complexity which are difficult to accommodate using conventional mathematic modeling methods. Although it is true that the total problem of digital filter optimization must be qualified as to the nature of the image, the imaging system, and the conditions of observation, an approach which we have found to be valuable in our own work, has been to limit the number of parametric variables so as to focus attention on those aspects of image quality which are most closely related to the parameters of the digital filter functions. At this point in our work, it was felt that additional levels of complexity would only serve to detract from the essential problem of determining the way in which digital filter characteristics relate to the detectability of lesions.

USE OF RESOLUTION PHANTOM FOR FILTER COMPARISON

Throughout this work we have utilized scans of a rather simple phantom which has been designed to emphasize three fundamental aspects of digital filter performance. The three quantitative criterion which can be measured in relative terms using this phantom are the degree of resolution enhancement, the amount of smoothing, and the tendency to produce artifacts. The configuration of this phantom is illustrated in Figure 1 and it has the immediate advantage of being easily fabricated from items available in most Nuclear Medicine laboratories. It consists of a thin flood source of Co^{57} (122 keV) with .125 inch thick lead disks of various diameters placed on the top of the flood source at the positions indicated. The positioning of the lead disks used as mock lesions in this phantom into three straight lines has been done to facilitate the plotting of cross sectional cuts through the mock lesions. Care has been taken to place these mock lesions far enough apart to avoid a periodic composition which might be falsely enhanced by digital filtering. For the purpose of comparing linear-digital filters the resolution phantom was scanned on a 5 inch Magnascanner using the 2114-B collimator. The plane of the phantom was placed three inches from the bottom of the collimator and the image data was sampled using image elements with dimensions of 0.1 x 0.125 inch.

This phantom has been designed for the sole purpose of evaluating digital filter performance. Existing phantom designs commonly used for measuring imaging system performance such as the bar phantom, the star phantom, Hine's phantom or the liver phantom are generally unsuitable in this regard. A filter function should be assessed in terms of its ability to detect small circular defects on a broad uniform background. Therefore, the use of phantoms which consist of closely spaced, periodic features can lead to highly erroneous conclusions in terms of filter function performance.

No attempt has been made in the design of this phantom to duplicate or simulate such scanning conditions as organ thickness or scatter within the patient. It is simply intended to provide a relative measure of the resolution enhancement, smoothing, and artifact creation incurred by the use of various digital filtering techniques. Therefore, no conclusions can be derived from its use concerning the size of lesions which can be detected in actual clinical images. This phantom does offer two significant advantages in that it is easily constructed in those laboratories having the capability for digitizing scintigraphic images as well as being easily simulated by those workers having access to a computer but no available means for digitizing scintillation data. This ease of duplication should serve to facilitate the intercomparison of image processing methods used by various investigators.

Three quantitative measurements are obtained from processed scans of this resolution phantom for evaluation of filter function performance. The first indicator is count ratio as a function of the diameters of the mock lesions. Count ratio is defined as the quantity C_t/C_o where C_t is the drop in count rate at the center of the mock lesion as compared to the average count rate, C_o , in an annular area surrounding the mock lesion. Under the conditions created by the design of the resolution phantom, count ratio as defined above is identical to the target to non-target ratio as well as being essentially equivalent to the concept of contrast efficiency as defined by Rollo and Schulz.⁽¹⁵⁾ Contrast efficiency is defined as being the contrast achieved by a given system divided by the contrast which would be achieved by an ideal imaging system. Since the thickness of the lead slugs which simulated the mock lesions of the resolution phantom used in this study very nearly attenuates all the radiation above the lesion locations, the contrast for an ideal imaging system would be approximately equal to 1.0. All of the above mentioned concepts (count ratio, target-to-non-target ratio, contrast efficiency, and contrast) are generally considered to be indicative of the detectability of an abnormal area when viewed by a human observer against a uniform background. This is not to say, however, that probability of detection is a simple function of contrast. When dealing with digitally filtered images, it is equally important to operate on the image in such a way as to maintain a uniform background in those portions of the image where there are no abnormalities. Count ratios for the six filters are shown in Figure 4.

This brings us to the second indicator of filter performance which is the equivalent noise bandwidth of the filter function. Equivalent noise bandwidth is a measure of the statistical smoothing provided by a given filter function. For the situation at hand, in which the statistical fluctuations originate from a non-uniform Poisson process, there are two equivalent methods to determine the equivalent noise bandwidth of a filter function. The method based on the integral of the frequency domain filter function is defined in Figure 5. Another equivalent method is based on Parseval's theorem. Noise power is proportional to the mean squared deviation about the average count rate over the uniform portion of the resolution phantom. Therefore, equivalent noise bandwidth becomes an approximate function of the mean square deviation (excluding lesions) about the average count rate before and after filtering the resolution phantom images. It is true, therefore, that the filter function which displays the lowest equivalent noise bandwidth has achieved the minimal least squared error -- for an image with no distinguishing features. This should not be confused with the minimum least squared error criterion proposed by Hart and Farrell.⁽¹⁷⁾ It must be kept in mind, therefore, that maximum count ratio and minimum noise bandwidth are not compatible objectives.

The third indicator is the least understood of the three but it is not any less important. This facet of filter performance concerns the tendency of the filter to produce artifacts. If we consider artifacts to be ringing phenomenon associated with areas of the image where there are sudden discontinuities in the image data, then a good measure of the tendency of a filter function to produce artifacts is the percentage overshoot in the filter output when the input function consists of the sharpest discontinuity the system is able to resolve (i.e., a unit step function convolved with the point spread function). This situation is illustrated in Figure 6. The tendency for the output of a filter function to overshoot for a step function input is not at all unrelated to the two indicators already discussed. For this reason, a pragmatic approach to filter function optimization must be achieved at the expense of a compromise between all three of these indicators. Thus, one might expect that an acceptable filter function would achieve maximum count ratios while minimizing noise bandwidth and at the same time maintain the percentage overshoot within acceptable limits.

RESULTS OF LINEAR FILTER STUDY

In order to evaluate the usefulness of linear digital filters for processing clinical scintigraphic images an intercomparative study was undertaken. This study was

designed to provide answers to two fundamental questions concerning some of the previously proposed digital methods for processing static images. These questions were: 1) Is there any validity associated with attempts to achieve resolution enhancement over and above smoothing of static images? 2) If resolution enhancement is realistically achievable, which of the many proposed techniques is most suitable for enhancement of routine clinical data? It was our feeling that the answers to these questions could best be obtained by processing a variety of clinical and resolution phantom data using a selected set of filter functions under identical conditions.

The group of filters selected for intercomparison consisted of six functions (one "representative" smoothing filter and five smoothing/enhancement filters). The six filter functions all have closed form frequency domain expressions which are plotted in one-dimension in Figure 2. The "representative" smoothing function used here corresponds to a compromise between a "box-car" function and a Gaussian function. It is essentially a triangular or Simpson weighted spatial smoothing function (with the exception that the undesirable frequency domain side-lobes have been truncated). The five smoothing/enhancement filters which were studied have been implemented in such a way that the peak values of the frequency domain function were all about 1.5 times the zero frequency values. That is to say that all five smoothing enhancement filters were set up to provide the same amount of high frequency emphasis but not necessarily at the same frequency.

It is interesting to note that this selection of parameters causes these five filters to take on the appearance of a family of filter functions. This feature of the study is coincidental and is surprising in light of the fact that the originators of these filters used very different rationales in deriving the functions used in this study. The details of the derivations of these five filters will not be discussed here as they are set forth elsewhere.⁽⁷⁻¹²⁾ It should be mentioned, however, that the originators of these functions were all seeking to develop filter functions which would provide a two-fold benefit in terms of increased resolution as well as improvement in the statistical reliability of digitally processed scintigraphic images. In order to understand the results of the intercomparative study, it must be kept in mind that the techniques of resolution enhancement and smoothing are fundamentally at odds when it comes to operating on "noisy" data. That is, filter functions which operate on the "signal" portion of a "noisy" image in such a manner as to improve resolution must be applied in moderation so as to avoid a significant amplification of the noise level. An analogous situation exists in radar signal detection theory where it is well known that signal processing procedures which increase the probability of detection must be traded-off against a coincident increase in the probability of false detections.

The results of processing a scan of the resolution phantom shown in Figure 3 show improved detectability of the smaller mock lesions. Examination of these filtered outputs reveals that the smaller mock lesions (.25 and .375 inches in diameter) are substantially more visible in those images which were processed with smoothing/enhancement filters. This fact is made more evident in the plots of count ratio versus lesion diameter shown in Figure 4. For a mock lesion diameter of .5 inches the count ratio is increased 50 to 75 per cent beyond the simple smoothing result by the five smoothing enhancement filters. The fact that this increase is rapidly approaching the point of diminishing returns as the bandwidth of the smoothing enhancement filters increases is revealed by the narrow range in which the count ratios for the four filters with the highest cutoff frequencies lie. The result is indicative of the fact that very little signal power is contained in the scan power spectrum at these higher frequencies (1.5 to 3.0 cycles per inch).

It is also evident from the results of Figure 3 that the higher count ratio values have not been achieved without detriment. There is an increasingly mottled appearance of the filtered outputs of those smoothing/enhancement filters having higher cutoff frequencies. This feature of Figure 3 reflects the larger magnitudes of the noise bandwidths of these filters. That is to say, the higher cutoff frequencies,

which are intended to give increased emphasis to the higher frequency "signal" power, are actually operating on more "noise" power instead. While, this fact is clearly evident from the frequency domain representation of the functions, it is unfortunately true that frequency domain analysis has played very little part in the bulk of work devoted to developing these filter functions.

The presence of artifacts in the filtered images of Figure 3 is most apparent near the edges of the resolution phantom. It is an intentional aspect of the design of the resolution phantom to present a sharp discontinuity at the edge in order to amplify the artifact producing tendencies of the filters. There is no tendency of the smoothing filter to produce artifacts which is reflected by the zero overshoot value for the smoothing filter given in Figure 6. The result of applying the Kirch-Brown filter to the phantom data shows faint artifact features which are most prominent near the corners of the phantom. This also is reflected in Figure 6 by the fact that the Kirch-Brown filter has the highest overshoot value. It is difficult to differentiate between the mottled appearance of low frequency noise and artifacts present in the images which have been filtered by those functions which have a higher cutoff frequency than the Kirch-Brown filter. It should be mentioned that, although the scan data at the edge corresponds to the sharpest discontinuity resolvable by the scanning instrument, this sort of edge function is seldom present in clinical data.

The six filter functions used in this study were also applied to routine clinical data. A typical right lateral liver scan is shown in Figure 7 following filtering with the six functions. The same sort of mottling which was apparent in the phantom data is present in the liver scan data which has been filtered by those functions with cutoff frequencies above 2.0 cycles per inch. The upper middle lesion which is not apparent in the unfiltered data, is readily visible in all of the filtered images.

CONCLUSIONS ABOUT LINEAR FILTERING OF STATIC IMAGES

The fact that the smallest lesions in the phantom scan as well as the liver scan are more readily apparent when processed by smoothing enhancement techniques serves to substantiate the validity of resolution enhancement as an achievable end -- within certain limits. The Kirch-Brown filter has been designed with these limits specifically in mind. That is, we have attempted to maximize count ratio while minimizing the noise bandwidth and maintaining the tendency to produce artifacts within reasonable limits. While it is true that the other smoothing enhancement filters studied here have parametric representations which allow the cutoff frequencies to be adjusted downward so as to reduce the mottling, in every case this would substantially reduce the peak value (and the resultant count ratio) of the filtered data. That is, all of these filters approach simple smoothing filters as the cutoff frequency is lowered. The common misconception shared by nearly all of the investigators proposing these other filter functions at the time of their development concerns the underlying role which least-squared error criterion or convergence to the original noiseless isotope distribution has played in their work. Charles Metz points out in his thesis⁽⁹⁾ that matched filtering in an effort to minimize least squared error can mistakenly "throw out the baby with the bath water."

Furthermore, we do not consider the linear filter function developed by ourselves during the course of this study to be the ultimate answer to processing clinical images. The developmental work which led to this filter design served better as a stimulus for further filter development rather than as a stopping point. One very disturbing fact that became apparent during the course of this intercomparative study was that certain types of clinical scans (specifically, brain scans) were unsuitable subjects for processing by any of the smoothing enhancement functions of Figure 2. It was generally necessary to decrease the cutoff frequency below 1.5 cycle per second to produce an output image without the characteristic mottled appearance that is associated with a higher than optimal noise bandwidth.

Another enlightening development concerned the widely accepted (but

unsubstantiated) importance attached to the use of the inverse of the modulation transfer function (IMFT) as a basis for deriving smoothing enhancement filter functions. Some informal experiments which involved replacement of the IMFT in the Kirch-Brown filter with a logarithmic function proved to produce better results. At this point, it became apparent that the concept of convergence to the noiseless isotope distribution is not necessarily synonymous with resolution enhancement. The considerations which have been summarized here led to the development of an adaptive filter for smoothing and enhancement of static images which will now be discussed in detail.

ADAPTIVE PROCESSING OF SCINTIGRAPHIC IMAGES

It stands to reason that, if the problems associated with smoothing and enhancement of images are fundamentally at odds, then it might be a good idea to get the statistical fluctuations present in an image well under control prior to the application of any sort of enhancement function. Our approach in doing so led to the development of an adaptive technique for smoothing each individual image in such a way as to take into account both the statistical nature and the frequency composition of each particular image. This technique uses the frequency domain concept of "average random noise level" to delineate between those Fourier coefficients which are primarily noise and those which can be considered as part of the signal portion of the image. The method for performing adaptive smoothing follows. First, the average random noise level is computed either from the total counts present in the image or by averaging the magnitudes of all the Fourier coefficients above the resolution limit of the imaging device. The magnitudes of all of the Fourier coefficients of the frequency transform of the image are then compared to the magnitude of the average random noise. If the coefficient is larger than the random noise level it is retained. If it is smaller, it is set to zero. If higher limits of statistical confidence are required, then a factored value (greater than one) of random noise level can be used to discriminate between noise and signal. It is interesting that the perimeter formed in the frequency plane by this sort of smoothing is generally very non-circular in shape. It is our view that the departure from Hankel smoothing functions brought about by this adaptive smoothing technique makes the adaptive technique less susceptible to the production of artifacts.

Once the frequency transform has been adaptively smoothed, it is now possible to approach the possibilities for resolution enhancement in a somewhat more innovative manner. Of course, one could simply apply the IMFT to the remaining coefficients following the adaptive smoothing operation. Our results using this adaptive approach have been proven to be very similar to those shown in Figure 3 with the only noticeable difference being absence of artifacts around the edges of the phantom. More impressive enhancement results were achieved by applying the logarithmic function plotted in Figure 8 to the adaptively smoothed data. A family of logarithmic curves were studied with the intention of determining the limits of resolution enhancement by means of this technique. The results of processing the resolution phantom with the Kirch-Brown linear filter and the adaptively enhanced non-linear filter are shown in Figure 9. Our choice has been to use the logarithmic function which peaks to a value of 3.0 at 2.0 cycles per inch. The resultant count ratio for the linear filter and the adaptive filter are plotted in Figure 10. This plot shows a 50 per cent increased count ratio achieved by the adaptive filter beyond the linear result for a mock lesion .5 inches in diameter. It is interesting to note the adaptive technique seems also to have reached a point of diminishing returns similar to that shown in Figure 5.

Another technique which we have found to be helpful in processing static images is called homomorphic filtering. This technique can be applied with linear techniques as well as with the adaptive technique. It consists simply of taking the logarithms of the data prior to performing the smoothing/enhancement operation and finally exponentiating the data following the filtering operation. It serves to allow somewhat independent control of the contrast of the fine detail present in the image independent

of control of the contrast of the broad expansive features of the image.

Finally, throughout the discussion so far no consideration has been given to the problem of human interpretation of image data. A technique which we have found helpful in this regard is to adjust the gray shade linearity in such a manner as to achieve a maximum entropy condition. From information theory we know that the maximum information is transmitted through a channel when all of the possible codes are equally probable. This condition can be created for each individual image by adjusting the gray-shade boundaries in such a way that there are equal numbers of image elements in each shade of gray.

Results of processing the same liver scan used earlier with the linear filter, the adaptive filter, the homomorphic filter, and finally the maximum entropy condition are shown in Figure 11.

DIAGRAM OF PHANTOM DIMENSIONS SHOWING DIAMETER AND POSITION OF .125" THICK LEAD SLUGS

Figure 1. Diagram of resolution phantom used for intercomparison of digital filter function.

Figure 2. One dimensional plots of the six linear digital filter functions which were compared in this study.

RESOLUTION PHANTOM (COUNT DENSITY = 2400 COUNTS/cm²)

COMPARATIVE RESULTS OF FILTERING RESOLUTION PHANTOM
(COUNT DENSITY = 2400 COUNTS/cm²)

BUDOTH (H=0)

KURO (H=0.025)

YAMAKA - MINAMI (PO) (H=0.05)

METZ (H=4)

KUNT-MEDER (H=7)

NABAI - NIKURA (H=10)

Figure 3. Results of processing a scan of the resolution phantom shown in Figure 1 using the six linear filter functions shown in Figure 2.

Figure 4. Plot of the resultant count ratios achieved by the six filter functions shown in Figure 2.

NOISE BANDWIDTH AND NOISE REDUCTION

B = THE RADIUS OF A CIRCULAR FILTER OF AMPLITUDE 1.0 WHICH WILL HAVE THE SAME NOISE POWER OUTPUT AS $F(\vec{\omega})$ FOR WHITE NOISE INPUT

$$B = \frac{\int_{-\frac{\omega_s}{2}}^{\frac{\omega_s}{2}} |F(\omega_x, \omega_y)|^2 d\omega_x d\omega_y}{4\pi^3}^{1/2}$$

R = THE NOISE POWER REDUCTION IN DECIBELS

$$R = 10 \cdot \log_{10} \left(\frac{\text{NOISE POWER OUT OF FILTER}}{\text{NOISE POWER INTO FILTER}} \right)$$

FILTER NOISE REDUCTION CHARACTERISTICS		
FILTER TYPE	B (CYCLES/INCH)	R (DB)
SMOOTHING	1.26	-13.01
KIRCH-BROWN	1.46	-11.74
TANAKA-IINUMA ($\sigma_{\omega_c} = .956$)	2.09	- 8.60
METZ (I = 4)	2.65	- 6.56
HUNT-MEDER (J = 7)	2.96	- 5.62
NAGAI-IINUMA (K = 10)	3.36	- 4.50

Figure 5. Definitions and tabulated values of noise bandwidth (in cycles/inch) and noise reduction (in decibels) for the six filter functions shown in Figure 2.

FILTER RESPONSE TO THE PSF * UNIT STEP

$$\Delta \text{ SLOPE} = \frac{\text{MAXIMUM SLOPE OF FILTER OUTPUT}}{\text{MAXIMUM SLOPE OF PSF * UNIT STEP}}$$

FILTER RESPONSE CHARACTERISTICS			
FILTER FUNCTION	% O.S.	S.D.(INCHES)	Δ SLOPE
SMOOTHING	0.0	.35	0.98
KIRCH - BROWN	8.4	.78	1.16
TANAKA - IINUMA ($\sigma\omega_c = .96$)	6.9	.46	1.39
METZ ($i = 4$)	6.1	.33	1.50
HUNT-MEDER ($J = 7$)	3.9	.26	1.49
NAGAI-IINUMA ($K = 10$)	1.2	.23	1.46

Figure 6. Definitions and tabulated values of percentage overshoot, settling distance (in inches), and the change in slope for the responses of six filter functions shown in Figure 2 to a unit step function convolved with the point spread function of the collimator.

**RIGHT LATERAL LIVER SCAN
(COUNT DENSITY = 800 COUNTS/cm²)**

**COMPARATIVE RESULTS OF FILTERING RIGHT LATERAL
LIVER SCAN (COUNT DENSITY = 800 COUNTS/cm²)**

Figure 7. Results of processing a liver scan (right lateral view) using the six filter functions shown in Figure 2. This scan contains proved lesions in the upper and lower central areas as well as along the lower left (posterior) edge of the organ.

Figure 8. Plot of the linear Kirch-Brown filter function (F_{KB}) and the three logarithmic functions used to evaluate the adaptive smoothing/enhancement technique.

COMPARISON OF LINEAR AND ADAPTIVE FILTER
(COUNT DENSITY = 2400 COUNTS/mm²)

Figure 9. Results of processing a scan of the resolution phantom using the linear Kirch-Brown filter and the adaptive smoothing/enhancement functions plotted in Figure 8.

Figure 10. Plots of the count ratios achieved by the filter functions shown in Figure 8.

Figure 11. Results of processing the same liver scan shown in Figure 7 using the linear, adaptive, homomorphic, and maximum entropy techniques.

CONSIDERATIONS OF IMPORTANCE IN ANALYZING DYNAMIC SCINTIGRAPHIC DATA

The development of hardware and software systems for handling dynamic scintigraphic data is presently emerging as an area of fundamental importance in Nuclear Medicine. Several dedicated minicomputer systems have been developed which feature extensive capability for accumulation, storage and display of dynamic scintigraphic data from Anger cameras. In general, these systems have very little provision for analyzing dynamic scintigraphic data beyond the ability to define areas of interest and extract corresponding time-activity curves. Our preliminary work in analyzing such time-activity curves has led to the development of an interactive, CRT oriented, curve analysis program. This program makes it possible to derive quantitative diagnostic information from these curves with minimal effort on the part of the clinician. The PDP-12 system referred to earlier has served as a developmental system for implimenting this curve analysis program and it is our feeling that programs of this type will greatly increase the usefulness of dedicated minicomputers in Nuclear Medicine.

Dedicated minicomputers which do not feature extensive curve analysis capability are little more than data management devices. Prior to the availability of computer/camera systems, diagnostic information had to be extracted from dynamic scintigraphic data by visual identification(21,22) or intermediate storage of data for later analysis on a larger machine.(23) We feel that an integrated data management and analysis system offers the sort of efficiency which will greatly improve the effectiveness of digital computers for routine clinical applications. At the same time, the flexibility we have maintained in our system makes it an invaluable tool for investigative and developmental work in deriving new diagnostic protocols.

STRUCTURE OF CURVE ANALYSIS PROGRAM

The curve analysis program used on our PDP-12 system has been adapted from the spectral analysis system (called "RUFUS") written by David Stern of the Laboratory for Atmospheric and Space Research, University of Colorado, Boulder, Colorado. The system is heavily display oriented enabling the user to get a much better "feel" for his data than is otherwise possible. Five-hundred and twelve point spectra or vectors are the basic data buffer and are stored in six registers in core. These spectra may be:

added and subtracted

multiplied and divided

integrated and differentiated

smoothed

edited

shifted

correlated and fitted with exponentials or straight lines

plotted

Fourier transformed and deconvolved.

The "language" of the curve analysis program consists of a large number of basic commands that may be entered manually or run under program control. A limited number of floating point and integer variables (not more than 26) are available for arithmetic, indexing and statement labels.

Minimum configuration for the program is a PDP-12 with 8-K core and two Linc Tape drives. Optionally supported peripherals include the KW-12A real time clock, a Calcomp 565 plotter, a card reader, an LP-8 line printer and an RK08 disk. All devices are run under interrupt to maximize the time the display is active.

Six spectra may be stored in memory at once in Spectrum Registers 0 to 5. Spectra may be input to registers from Linc Tape unit number one, from the card reader in binary form or from the RK08 disk. Registers can be output to Linc Tape 1, the teletype, plotter line printer, or RK08 disk unit 11.

Spectra are stored on Linc Tape Unit 1 or disk unit 11 using two blocks per spectrum beginning on each even block. 256₁₀ or 450₁₀ spectra may be stored on one tape, for standard linc tapes or 896₁₀ block tapes respectively.

Instructions to the system are in the form of "commands" entered from the teletype. Commands are interpreted in three modes: 1) manually - commands are executed as they are typed; 2) automatically - commands are stored in a 1024 character buffer and executed under program control with variables, loops, etc.; 3) storage mode - commands are stored in the program buffer as they are entered from the TTY and normally executed. (This is similar to "FOCAL".)

Programs may be written using the text editing features of the "DIAL" system and loaded by the curve analysis program from either a named DIAL file or from the working storage area. Also, programs can be saved on previously existing DIAL source files. These programs are completely compatible with the DIAL text editor.

The Display: Every effort has been made to make the display as interactive and informative as possible. One or two spectra may be displayed at once, selected directly by the operator or automatically as the result of a command. The x-axis can be expanded from 512 points to two points across the entire screen depending upon the setting of analog channels 1 and 5. The vertical separation between the two displayed spectra or the position of one spectra is controlled by knob 4. Full screen range vertically is 512 normally, or if sense switch 2 is depressed all y-values are divided by four before they are displayed giving a range of 2048.

The register number for each spectrum is displayed about 1 1/2 inches above the x-axis at the left of the screen.

There are four variable features of the display, one is the vertical spacing or position controlled by knob 4. The other three are the cursor (knob 0) and the high and low expand limits, "HIEXPD and LOEXPD" (knobs 1 and 5). These three features are used by various commands for parameters or may be set under program control by other commands.

The "cursor" set by knob 0 is the one intensified point in each register at the same x-value. This x-value is determined by the setting of knob 0. The y-value of the cursor is the value of the displayed register at that point.

The two vertical dashed lines are LOEXPD and HIEXPD from left-to-right. If these ever cross, i. e., LOEXPD greater than HIEXPD, the display will go blank until this is corrected. If sense switch 1 is set to one, the portion of the displayed registers between LOEXPD and HIEXPD is expanded to fill the screen horizontally. The portion between LOEXPD and HIEXPD may be varied continuously. If sense switch 1 is zero, the full 512 points will be displayed.

Sense switch 0 determines whether one or two spectra are displayed. If it is zero, two spectra are displayed. If it is one, only one spectrum is displayed with this additional feature: four numbers are displayed along the bottom of the screen. They are from left to right: 1) LOEXPD x-value, 2) cursor x-value, 3) cursor y-value and 4) HIEXPD x-value. When SNS 0 = 1, only the "UPPER DISPLAY" register is

displayed. When SNS 0 = 0, both the "UPPER DISPLAY" and "LOWER DISPLAY" registers are displayed.

SNS 3 controls the input and output buffer displays which appear above the numbers on the bottom of the screen.

A summary of the display controls are as follows:

SENSE SWITCH 0 - dual or single register display

- 1 - normal or expanded display
- 2 - normal or y/4 scaling
- 3 - no buffer display or both buffers displayed
(usually set to 1)

ANALOG KNOB 0 - cursor x-value

- 1 - LOEXPD x-value
- 4 - y position
- 5 - HIEXPD x-value.

If when the program is first loaded and there is no display, rotate knob 1 counter clockwise and knob 5 clockwise until the display returns.

Arithmetic: Twelve bit signed integers are used for spectrum data values, allowing a range of -2048_{10} to $+2047_{10}$. Y-values above the top of the screen or below the bottom will wrap around the screen.

Variables: The programming system allows 26_{10} variables ("A" to "Z") that may be integers, program labels or floating point variables. A variable may be used wherever a constant is used and may be preceded by a minus sign. A letter may have only one of the three uses described above in any one program. Floating point variables occupy three locations starting with the name of the variable. The floating point variable "X" uses locations "X", "Y" and "Z". These locations may not be used for anything else. Caution must be used to insure that a floating point variable does not destroy integer variables or labels.

Command Format: Each command has a one or two letter name followed optionally by one or more parameters and terminated by a carriage return.

The character (RUBOUT) will erase the entire line and echo "<<". The maximum line length on input is 30 characters. Legal characters are the letters A - Z, the digits 0 - 9, the minus sign -, rubout, carriage return and the character "CNTRL/R".

AN EXAMPLE OF USING THE CURVE ANALYSIS PROGRAM TO ANALYZE DYNAMIC SCINTIGRAPHIC DATA

A new method of determining atrial and ventricular volumes and assessing ventricular function by dynamic scintigraphic imaging of the heart serves to illustrate the usefulness of the curve analysis program. These studies are performed using an Anger camera which has been interfaced to a PDP-12 digital computer capable of accumulating scintigraphic image data at the rate of ten frames per second. With the patient in the RAO position a bolus of 6 mCi of Technetium as pertechnetate, followed by a 10 cc saline flush, is delivered through a Swan-Ganz catheter placed in the pulmonary wedge position. The catheter is then withdrawn to the superior vena

cava position and a second injection is made. The sequence of images collected following these two injections provide excellent visualization of the four heart chamber contours. The computer is then used to define areas-of-interest coincident with the four heart chambers and the time-activity washout curves are extracted from the sequential image data and analyzed.

Forward cardiac output is obtained by indicator dilution technique. Ventricular ejection fraction and end-diastolic volume are obtained by two methods, which correlate well: 1) analysis of peak-to-valley ratios of ventricular activity over several cardiac cycles, and 2) an exponential fit of the ventricular washout curve. End-diastolic volume is then solved for as a function of the exponent of decay and cardiac output. A measure of tricuspid and mitral regurgitation can also be quantitated by compartmental analysis of the data using the appropriate atrial-ventricular model.

In addition, the data accumulated following the vena cava injection allows an independent measure of left ventricular ejection fraction by the technique of Van Dyke.⁽²⁴⁾ This new method offers the advantage of assessment of left ventricular function by a relatively non-invasive technique, avoiding arterial catheterization.

SOME PRELIMINARY CONCLUSIONS REGARDING THE USE OF THE CURVE ANALYSIS PROGRAM TO ANALYZE CARDIAC FLOW DATA

The technique outlined in the preceding section has been found to provide excellent washout studies of both the right and left heart in approximately 70% of the 23 patients studied to date. It remains true, however, that no matter how good a bolus is presented to the right and left atria of the heart, several problems occur which degrade the washout data extracted from the areas-of-interest coincident with the four heart chambers. One major problem concerns cross-talk due to the presence of radionuclide in anatomic structures surrounding the heart. In the case of right heart studies this is primarily due to pulmonary circulation and in the case of left heart studies it is due to descending aorta, bronchial circulation and mitral flow. We have found that a background area can be defined for each of the four heart chambers which surrounds each heart chamber in the annular fashion of Van Dyke⁽²⁴⁾ (taking care to avoid the major vessels into and out of the heart). Figure 12 shows the washout curves and associated time-variable background curve for each of the four heart chambers obtained by the method outlined in the preceding section. Using the curve analysis program the background curves are scaled up or down to match the tail of the washout curve. When the time-variable background is subtracted from the washout curve the curves shown in Figure 13 result. The exponential curve fitting capability of the curve and analysis program has been used to fit these corrected curves and the result has been superimposed on top of the corrected curves. The exponential time constant obtained in this fashion can be used to determine such parameters of cardiac function as ejection fraction and end diastolic volume in the absence of regurgitation.

Several determinations can be made from these curves. Quantitatively, two things are apparent. First, the consistent rise in the right ventricular curve following systolic shows a slight amount of regurgitation back into the right ventricle. This is most likely from the pulmonary artery since there is not any indication of right atrial regurgitation. Second, the very slow washout from the left ventricle (in the absence of regurgitation) indicates a dyskinetic L. V. aneurysm. Quantitatively, the most easily obtained parameter of cardiac function is ejection fraction. Using the curves shown in Figure 13 the ejection fractions for the right atrium, right ventricle, left atrium and left ventricle are respectively .425, .667, .556, .273. The very low value of .273 (normal = .65) is typical of the diagnosis of L. V. aneurysm.

While the example given above illustrates only a small part of the abilities of the curve analysis program to operate on curves, we feel that the other analytic components of the program will ultimately lead to better understanding of and methods for quantification of cardiac flow studies. We are currently exploring techniques for

Figure 12. Photographs taken from the face of the CRT display while using the interactive curve analysis program to analyze cardiac flow studies. Each plot shows the washout curve (upper curve) along with the time-variable background (lower curve) for that particular chamber.

Figure 13. The same curves plotted in Figure 12 following subtraction of the time-variable background from the washout curve. The fitted exponential is shown superimposed on each curve.

modeling and quantitating regurgitation which are heavily dependent on the ability of the curve analysis program to perform a wide range of tasks. These requirements range from the ability to simply read points off the curve at the times which correspond to systole and diastole (for beat-by-beat analysis) to the ability to deconvolve the ventricular activity curves with respect to the atrial activity curves (to compensate for late entry of the radionuclide from the atrium). While we are not proposing a method for quantitating regurgitation at this point, we are reasonably certain the curve analysis program already contains the basic building blocks which will be necessary to solve this problem.

ACKNOWLEDGMENT

We wish to gratefully acknowledge the motivation and participation of Doctors P. P. Steele,* M. E. Mathews,* D. H. Davies,* and D. Van Dyke,** who have devised and implemented the wedged pulmonary catheter technique for performing the cardiac flow studies described herein.

*Department of Cardiology, Veterans Administration Hospital, Denver, Colorado

**Donner Laboratory of Medical Physics, University of California, Berkeley, California

BIBLIOGRAPHY

1. D.W. Brown, "Digital Computer Analysis and Display of the Radioisotope Scan," J. Nucl. Med., 5, 802-806 (1964).
2. H. Schepers and C. Winkler, "An Automatic Scanning System Using a Tape Perforator and Computer Techniques," Medical Radioisotope Scanning, Vol. 1, 321-329 (Vienna, IAEA, 1964).
3. A.C. Sprau, W.N. Tauxe, and D.W. Chaapel, "A Computerized Radioisotope Scan-Data Filter Based on a System Response to a Point Source," Mayo Clin. Proc., 41, 585-598 (1966).
4. S.M. Pizer and H.G. Vetter, "The Problem of Display in the Visualization of Radioisotope Distributions," J. Nucl. Med., 7, 773 (1966).
5. D.E. Kuhl and R.Q. Edwards, "Digital Techniques for On-Site Scan Data Processing," Fundamental Problems in Scanning, 256-260 (Springfield, Ill., Charles C. Thomas, 1968).
6. W.J. Mac Intyre and J.H. Christie, "A Comparison of Data Averaging of Radioisotope Scan Data by Photographic and Dimensional Computer Techniques," Medical Radioisotope Scintigraphy, Vol. 1, 771-781 (Vienna, IAEA, 1968).
7. D.W. Brown, D.L. Kirch, T.W. Ryerson, et al, "Computer Processing of Radioisotope Scans Using the Fourier and Other Transformations," J. Nucl. Med., 12, 287-291 (1971).
8. E. Tanaka and T.A. Iinuma, "Approaches to Optimal Processing in Radioisotope Imaging," Phys. Med. Biol., 15, 683-694 (1970).
9. C.E. Metz, A Mathematical Investigation of Radioisotope Scan Image Processing, Ph.D. Thesis (University of Pennsylvania, 1969).
10. W.A. Hunt, H.G. Meder, P. Pistor, et al, Digital Processing of Scintigraphic Images, Heidelberg Scientific Center (IBM), Technical Report 70.03.001 (1970).
11. W.A. Hunt, H.G. Meder, P. Pistor, et al, Optimum Sample Size in Digital Radioscintigraphy, Heidelberg Scientific Center (IBM), Technical Report 70.09.004 (1970).
12. T. Nagai, T.A. Iinuma, and S. Koda, "Computer-Focusing for Area Scans," J. Nucl. Med., 9, 507-516 (1968).
13. W.C. Dewey and W.K. Sinclair, "Criterion for Evaluating Collimators used in In Vivo Distribution Studies with Radioisotopes," Int. J. Appl. Rad., 10, 1 (1961).
14. R.N. Beck and P.V. Harper, "Criterion for Evaluating Radioisotope Imaging Systems," Fundamental Problems in Scanning, 348-379 (Charles C. Thomas, Springfield, Ill., 1968).
15. F.D. Rollo and A.G. Schulz, "A Contrast Efficiency Function for Quantitatively Measuring the Spatial-Resolution Characteristics of Scanning Systems," J. Nucl. Med., 11, 53-60 (1970).

16. L.J. Maroglio, L.G. Knowles, L.C. Kohlenstein, et al, "Improvement of Lesion Detection Performance with Image Processing," (abstract) J. Nucl. Med. 12, 381-381 (1971) .
17. R.W. Hart and R.A. Farrell, "Imaging Distribution of Radioactivity within the Human Body," Invest. Radiol., 3, 199 (1968) .
18. B. Gold and C.M. Rader, Digital Processing of Signals (McGraw-Hill, New York, 1969) .
19. J.F. Kaiser, "Digital Filters," System Analysis by Digital Computers (Wiley, New York, 1966) .
20. J.M. Mozley, "The Modulation Transfer Function for Scanners," Fundamental Problems in Scanning, 301-313 (Charles C. Thomas, Springfield, Ill., 1968) .
21. J.P. Kriss, G.S. Freedman, L.P. Enright, et al, "Radioisotopic Angiocardiography," Rad. Clin. N. Amer., 9, 369-383 (1971) .
22. B.L. Zaret, H.W. Strauss, P.J. Hurley, "A Non-invasive Scintigraphic Method for Detecting Regional Ventricular Dysfunction in Man," New Eng. J. Med., 284, 1165-1170 (1971) .
23. J.R. Cox and K.B. Larson, Proceedings of the Workshop on Computer Processing of Dynamic Images from an Anger Scintillation Camera, Biomed. Comp. Lab., Washington University, St. Louis, Missouri (January 18-22, 1971) .
24. D. Van Dyke, H.O. Anger, R.W. Sullivan, et al, "Cardiac Evaluation from Radioisotope Dynamics," J. Nucl. Med. (to be published) .

COLLECTION, PROCESSING, AND DISPLAY OF DYNAMIC FUNCTIONS*

Nathaniel M. Alpert, Joan E. Correll,
Linda A. Deveau, and Marc-David Cohen

Physics Research Laboratory
Massachusetts General Hospital
Boston, Massachusetts

ABSTRACT

Dynamic function data from the M.G.H. positron camera, the scintillation camera and multi-probe devices are now being processed under our operating system, CROSS¹. The dynamic function processor is called DPS. DPS allows for unusual flexibility as a research tool while maintaining the convenience necessary for use in a clinical environment. A particularly important feature is the ability to process collected data while new studies are being acquired.

The processing routines and variable parameters, if any, along with the form of the input and output are specified by the user. Each routine performs one major function; however, up to sixteen routines can be chained to carry out complicated tasks. The DPS executive, core, and disk allocation are described with special consideration given to the optimization of disk allocation in our hardware configuration².

References:

1. Brownell, G. L., Burnham, C. A., Pizer, S. M., and Wilensky, S., "NUMEDICS - A Computer System for Processing Radioisotope Data from Multiple Sources," USAEC Conf-7-10425:51-60 (April, 1971).
2. Cohen, M., Deveau, L. A., Pizer, S. M., Harris, S. J., and Hamlin, G. A., Jr., "CROSS - An Operating System for Time-shared Scan Collection, Processing and Retrieval," USAEC CONF-7-10425:215-225 (April, 1971).

*This work was supported in part by PHS Grants: GM16712, GM00889, and AEC Contract AT(11-1)3333.

Collection, Processing and Display of Dynamic Functions

Nathaniel M. Alpert, Joan E. Correll, Linda A. Deveau,
and Marc-David Cohen

Physics Research Laboratory
Massachusetts General Hospital
Boston, Massachusetts

Introduction

In papers presented last year at this conference we described the general features of NUMEDICS¹, our hardware-software configuration for processing radionuclide data from multiple sources, and CROSS², an operating system for time-shared scan collection, processing and retrieval. Over the past year we have obtained a great deal of experience with both the hardware and software. The purpose of this paper is to outline how collection, processing and display of dynamic functions are integrated under CROSS. We also felt it would be of interest to workers in the field to discuss some of the recent improvements in CROSS along with programming considerations which we believe will further enhance its flexibility.

Programming Considerations in CROSS

The original file structure consisted of a linear directory of filenames, taking as much disk space as necessary for each file as it was collected or processed. There were three directories and file areas under CROSS control: one for unprocessed pictures, one for the processed pictures, and one for retrieved pictures. File deletion was impossible since there was only a pointer to the next available disk area.

This limitation was overcome for processed files by placing the processed area under the processor's control, and reallocating disk space so that each processor can use up to 500 K words ($K=1024$). The actual disk allocation is detailed in Table I.

The processed directory contains pointers to a list of 128 filenames in addition to a list of 512 words used to link the cells occupied by a single file. The unprocessed and processed file directories are similar, but instead of a linked list, the unprocessed file directory uses a bit map to represent the use or availability of a cell. An unprocessed file either takes up one or two cells. This is indicated by a file length bit which is part of the filename. In order to do only one disk write when saving a file at collection time, files which occupy two

CROSS DISK ALLOCATION

<u>SEGMENTS</u>	<u>DISK ALLOCATION</u>
2308	DPS PROGRAMS
10	DPS NAME DIRECTORY
8	DPS LINKED LIST DIRECTORY
10	DPS PARAMETERS
10	DPS PROGRAM ROLLOUT
4606	512-576 WORD CELLS FOR DPS FILES
65	ANGER CAMERA CORRECTION MATRIX
65	POSITRON CAMERA CORRECTION MATRIX
127	CROSS PROCESSOR ROLLOUT AREA
20	TAPE DIRECTORY AREAS
10	UNPROCESSED FILE NAME DIRECTORY
1	UNPROCESSED FILE BIT MAPS
3	RETRIEVAL FILE NAME DIRECTORY
1	RETRIEVAL FILE BIT MAPS
4224	128-2712 WORD CELLS FOR UNPROCESSED FILES
2784	32-5568 WORD CELLS FOR RETRIEVAL FILES
474	PLOT AREA
121	SLOW COLLECTION FILE AREA
1172	CROSS PROGRAM AREA

Table 1

cells are made to be adjacent. This new method makes deletion more practical and also allows for the collection of 64 x 64 nine bit pictures (henceforth referred to as 2K pictures).

The retrieval directory is also being changed so that files will be chained in chronological order. This will allow a single user to have the entire disk for retrieval, but when another user wants to retrieve processed files the disk area will then be divided equally forcing the first user to give up his oldest files.

More flexibility in disk allotment could be obtained by assigning disk space for various functions according to the instantaneous load on the system. In periods where user demand is low, more space could be available for individual users. All files would share the same disk area, but retrieved, processed and unprocessed files would each have their own directory. The disadvantage of this approach occurs when a user calls the DEC supplied programming system ("Advanced System") under CROSS, since it assumes certain disk space for its own use.

Collection of Dynamic Functions

Our collection devices fall into two categories. The gamma and MGH positron camera, which have high data rate capabilities, use the direct memory access feature of the Digital Equipment Corporation PDP-9 computer; whereas, dynamic function collection from discrete probe devices uses the program interrupt. Over the past year most collection has been done with the positron and gamma cameras. Both have similar collection programs for static and dynamic studies. In the static mode the beginning and end of collection is indicated by the user. The dynamic mode uses the computer's clock to time both the frame length and interframe time to a precision of 1 milli-second. The minimum frame time is 0.1 second. One of the most valuable features from a user standpoint is the ability to vary both the frame and interframe times as a function of the expected count rate. At present, we allow three sets of frame times.

Our experience has been that this approach is inflexible. At the Massachusetts General Hospital in the Physics Research Laboratory, the variety of dynamic function studies requires that the user be able to specify his individual collection routine. We are implementing a new scheme based on a utility program which will allow the user to create his own collection routines. The user will be allowed to specify collection on 2K or 4K frames and the option of stopping the individual frame collection after either a fixed count has been accumulated or a preselected time has elapsed. In the 2K collection mode it will be possible to specify up to 127 frame times and 126 interframe times. This collection information will be named by the user and stored for later use by the collection program. Stored with each frame will be the picture time, the interframe time and the number of counts in the frame.

Dynamic Processing System under CROSS

General

DPS, the dynamic function processing system under CROSS, has been developed over the past year and is now primarily a research tool. Since the optimum dynamic function processing for a specific study cannot always be predicted, DPS was designed to give the user the most freedom while processing under CROSS. Each routine under DPS performs only one function; but, routines can be chained to perform complicated processing tasks. As clinically valuable dynamic function processing packages are identified the individual routines can be combined for more efficient use.

All information necessary for the execution of processing routines is specified by the user. This includes not only the input files and the processing routines, but also the form of the output and any variable parameters. To provide the convenience of repeating or easily varying a processing procedure all parameters are stored by DPS and remain until changed or deleted.

The DPS Executive

When the user enters the DPS executive, the lower 8K of core is allocated as shown in Fig. 1. Each processing routine obtains its variable parameters, if any, from the routine variable table. On execution the routine variable table is written over and space is reallocated, as in Fig. 2, so that 6464 words of working core are available. After a routine is executed, the DPS loader reads the routine variable table back from the disk into core, reads in the next routine queued in the routine table, and begins execution. After the last routine in the queue is executed the core allocation shown in Fig. 1 is restored.

Because the DPS disk area is divided into 576 word cells, it must have its own handler to manipulate the disk input and output of the processing routines. It is dependent, however, upon calls to the CROSS disk handler. When the DPS disk handler is called it rolls out the processing routine and rolls in the DPS filename and linked list directories. After the disk transfer is completed it writes the directories back onto disk and rolls in the routine to continue processing.

Up to 16 routines and their order of execution can be specified to form a processing series. The DPS executive links each routine so that one is initiated immediately after another without further interaction by the user. Any processing routine variables defined by the user are accessible to the routines in a routine variable table. However, if the user doesn't assign a variable necessary within a routine, the routine itself sets that variable to a standard value.

DPS CORE ALLOCATION DURING EXECUTIVE CONTROL

FIGURE 1

DPS CORE ALLOCATION DURING ROUTINE EXECUTION

FIGURE 2

An important feature of the DPS executive is a utility package which is entered by the user typing a control character. It gives general information about the status of the DPS disk area and allows manipulation of the files stored there. Once in the DPS utility package, the user can clear the DPS disk area, list the names on the teletype of all files present, delete a file or sequential files, get a printout of the available space, tag a file or sequential files for CROSS storage, or call CROSS to store all tagged files.

Input to DPS

As implied above input files to DPS must be moved to the proper disk area. Transfer is accomplished by calling a locate and move routine which uses CROSS disk or DEC tape handlers. DPS assigns the transferred files a numerical position in the DPS name directory by which they can be referenced later. Once the input files are in the DPS disk area their order of use can be specified.

Output from DPS

The form of processed output can be assigned by the user. Options include any combination of output on teletype, display disk (64 brightness levels on a 128 x 128 raster), or on a CROSS storage area. Eventually hard copy will be available on a digital plotter.

Processing Programs under DPS

A library of dynamic function processing programs is being generated. Since DPS became available only during the past year, programs were written on a priority basis. More general modeling and simulation routines are planned. Currently, available routines include dynamic function extraction from up to eight picture areas, display disk plots of dynamic function, moment calculations, and dynamic background subtraction.

Summary

Experience over the past year with the dynamic function processing system DPS has demonstrated both its flexibility as a research tool and its utility in a clinical environment.

References

1. Brownell, G.L., Burnham, C.A., Pizer, S.M., and Wilensky, S., "NUMEDICS - A Computer System for Processing Radioisotope Data from Multiple Sources," USAEC Conf-7-10425:51-60 (April, 1971).
2. Cohen, M., Deveau, L.A., Pizer, S.M., Harris, S.J., and Hamlin, G.A., Jr., "CROSS - An Operating System for Time-shared Scan Collection, Processing and Retrieval," USAEC CONF-7-10425:215-225 (April, 1971).

INTERACTIVE PATTERN ANALYSIS AND CLASSIFICATION SYSTEMS*

Laveen Kanal

Computer Science Center
University of Maryland
College Park, Maryland 20742

ABSTRACT

This paper surveys a number of Interactive Systems and their application to a wide variety of patterns such as stock and commodity market patterns, images, speech, waveforms, palm-prints, medical data and alphanumeric handprinting. The paper comments on various aspects of alternative hardware and software implementations and discusses computational algorithms and mappings relevant to interactive analysis and classification of patterns.

April 1972

*This work was supported in part by the Directorate of Mathematical and Information Sciences, Air Force Office of Scientific Research, Air Force Systems Command, USAF, under grant AFOSR 71-1982 to the University of Maryland. The paper is to appear in a special issue of the Proc. of the IEEE on Digital Pattern Recognition, in September 1972.

INTERACTIVE PATTERN ANALYSIS AND CLASSIFICATION SYSTEMS*

By Laveen Kanal

SUMMARY

In problem-oriented pattern recognition investigations, the starting point is usually some samples of patterns. The question then asked is whether or not the recognition of patterns such as these can be automated, and if it can, how complex a machine will be needed [Kanal and Harley (196)].

To go from these initial vague questions to an actual pattern recognition system involves a series of refinements and formalizations concerning the deterministic, probabilistic or mixed structure that we can infer about the patterns; what level of performance we should strive for; what competing design approaches we think are worth considering and what manner of implementation is relevant.

The first step in this process is Pattern Analysis, which is most important in guiding subsequent steps. This is one aspect of Pattern Recognition which is greatly enhanced by Interactive Systems. Furthermore, in order to understand the variability in relatively unconstrained data and to come up with solutions, we need to study the detailed peculiarities of a very large data base. Until recently, extensive experimentation on large amounts of data was out of the question, without spending an inordinate amount of time and money. We have only to think of the patience of various great scientists to remind ourselves that the latter approach has been the one followed historically. Today very few institutions and almost no commercially-oriented engineering organizations interested in pattern recognition allow their scientist-engineers that luxury of time. But for practical pattern recognition problems, unless there is an easy way of trying out competing techniques on a large amount of data, we are usually stuck with one or two approaches.

Also, we are no longer content to do what was often done a decade ago, i.e. jump to a hardware implementation of a particular learning machine, an optical spatial filtering system, etc. It is true that one can go a long way, using a problem-oriented approach and a theoretical analysis of requirements, toward rejecting candidate solutions, as was shown [Harley et al (1968)] for the problem of detecting and locating objects of military interest in aerial reconnaissance photography. That analysis showed what invention was needed before optical spatial filtering techniques would be practical for automatic target detection,

and made apparent the electronic methodology and classification design approach that would serve the problem needs. Nevertheless, that type of analysis is limited to certain overall aspects of a problem.

The desirability of an automated interactive approach to pattern analysis and classification logic design can be summarized:

(1) No single model exists for all pattern recognition problems and no single technique is applicable to all problems. Rather what we have in pattern recognition is a bag of tools and a bag of problems. We want a flexible tool kit and as painless and as quick a way of trying out a variety of tools that appear relevant to the problem at hand.

(2) Feature definition and extraction, and pattern classification, are best examined via trial and evaluation. The boundaries between feature selection and classification are not sharp. We need feedback between feature selection, logic design, classification, and testing. And, we need many iterations of the feedback process. This is naturally an interactive operation. We cannot preprogram a batch mode operation to go through all the possibilities--it would be too time consuming and costly. So we turn to an on-line and interactive mode of operation.

(3) Most data analysis techniques are trying to answer questions about the structure of the data in a high dimensional space. Interactive pattern analysis is a way of allowing examination of structure. (Humans are superior in recognizing structure of certain types, especially clusters, where automatic clustering routines are often thrown off by outliers between clusters.) However, to make this examination of structure feasible we must arrange for special ways of displaying relationships between data elements.

(4) In many pattern recognition problems, the human is the standard; we should put him in charge of the design and evaluation process but help him with automation in an interactive mode. For good interaction, the lag between the initiation of some control function and the completion of the requested action should be short enough that it will not interrupt his train of thought.

It is now recognized that the key to pattern recognition problems does not lie wholly in learning machines, statistical approaches, spatial filtering, heuristic programming, formal linguistic approaches or any other particular solution which has been vigorously advocated by one or another group during the last decade and a half as the solution to the pattern recognition problem. Rather what we have in pattern recognition is a bag of problems and so we must have a versatile tool kit and gain experience and judgement in the use of the various tools, be they statistical, linguistic, heuristic, etc. We also need to gain experience and judgement concerning the generality of various approaches. For

example, [Lederberg and Figenbaum (1968)] in commenting on the pattern recognition aspects of their program for the mechanization of inductive inference in organic chemistry, are able to say "In sum, we find that the development of this program has not encountered very much that is fundamentally new in principle. Problem solving in this field has much the same flavor as the solutions already adduced for chess, checkers, theorem proving, etc."

We would like to have a quick, flexible way of analyzing sample patterns, trying out various tools from our tool kit and only when we feel reasonably sure that a certain algorithm or approach is more promising than others, consider implementing it as hardware or firmware. This exploratory pattern analysis and evaluation of competing tools is greatly enhanced by automation of known procedures [Tukey (1962)] so that they can be applied in a routine fashion. It is further enhanced by interactive systems. Standardizing the testing of our hypotheses concerning data requires that we incorporate a variety of statistical and nonstatistical procedures; we must understand their theoretical properties and gain experience regarding their practical limitations and capabilities.

In our paper [Kanal 1972] from which this presentation has been summarized, we survey approximately twenty representative Interactive Pattern Analysis and Classification Systems (IPACS) and their applications. Included are a most fully developed commercial application of an interactive graphic pattern analysis system for the analysis of stock and commodity market patterns, and systems for multivariate statistical analysis and classification of vector data using clustering, discriminant analysis and related statistical routines in an interactive-graphics environment. Also included are "front-end" systems for interactive image, speech and waveform processing. One of the IPACS applications discussed in the paper, which may be of special interest to a medical audience, concerns analyses of dermatoglyphic (palm-print) patterns to determine whether there is sufficient information in the palm to discriminate among a variety of diseases; in particular, these studies of dermatoglyphic patterns have tested the validity of some conjectures on the familial transmission of Down's syndrome.

The paper [Kanal 1972] examines design considerations for interactive pattern analysis and classification systems and discusses various approaches to the realization of the capabilities desired in a system. It also presents some details of the computational capabilities of representative systems.

ACKNOWLEDGEMENT

This work was supported in part by the Directorate of Mathematical and Information Sciences, Air Force Office of Scientific Research, Air Force Systems Command, U. S. A. F. under grant AFOSR 71-1982 to the University of Maryland.

REFERENCES

- *Kanal, Laveen (1972), "Interactive Pattern Analysis and Classification Systems - A Survey and Commentary", Tech. Report, Grant AFOSR 71-1982, 111 pages, 252 references, University of Maryland Computer Science Center, College Park, Maryland, 20742; Scheduled to appear in 1972 in a special issue, of the Proc. of the IEEE, on digital pattern recognition, edited by Leon Harmon.
- Kanal, L., and T. J. Harley, Jr. (1969), "The complexity of patterns and pattern recognition systems", Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio, Technical Report No. AMRL-TR-69-62, November 1969.
- Harley, T. J., L. N. Kanal, and N. C. Randall (1968), "System considerations for Automatic Imagery Screening", Pictorial Pattern Recognition, G. C. Cheng et al (eds.), Washington, D. C.: Thompson Book Co., pp. 15-32, 1968.
- Lederberg, J. and E. A. Feigenbaum (1968), "Mechanization of Inductive Inference in Organic Chemistry", Carnegie Tech. Symposium on Cognition, in Formal Models of Human Judgment, B. Kleinmuntz (ed.), New York: John Wiley, 1968.
- Tukey, J. W. (1962), "The Future of Data Analysis", Ann. of Math. Stat., vol. 33, no. 1, pp. 1-67, March 1962.

Abstract

Instrumentation for Pattern Recognition and Associated Software

by

Robert S. Ledley and Louis S. Rotolo

In this paper we describe the biomedical pattern recognition system of the National Biomedical Research Foundation. The hardware portion of the system consists of a high-resolution flying-spot scanner (FIDAC), a vidicon scanning system with associated silicon video memory system (VIDIAC), a mechanical scanner of very great resolution (DRIDAC), a microscope scanning system (SPIDAC), and an interactive communication system (MACDAC). The associated software systems used in the operation of the above hardware systems and in the automatic pattern recognition consist of FIDACSYS, which supervises input of the image into the computer from the scanning devices and accomplishes overall picture manipulation, and the SYNTAXSYS and BUGSYS pattern-recognition languages developed at the Foundation. Finally, a system (DOCS) is described which is used for on-line computer-console interaction with disk memories for the evaluation and display of large masses of data.

April 21, 1972

National Biomedical Research Foundation
Georgetown University Medical Center
3900 Reservoir Road, N. W.
Washington, D. C. 20007

Instrumentation for Pattern Recognition and Associated Software *

The pattern recognition system of the National Biomedical Research Foundation consists of a number of scanners, display consoles, and interactive devices. The laboratory is centered around an IBM 360/44 computer (owned by the Foundation), which is used on a "dedicated" basis. The figures show the organization of the specialized pattern-recognition equipment of the laboratory (see Figs. 1, 2, and 3).

All of the equipment (except the IBM 360/44 computer) is of our own design and construction; either it was not available from commercial electronic manufacturers or its requirements exceeded the specifications of available apparatus. All of the scanners and interactive displays are integrated into a single configuration through the Pattern Recognition Control System. The SPIDAC (Specimen Input to Digital Automatic Computer) consists of an automated microscope with a motorized stage which is under computer control in both the x and y directions; it includes an instantaneous automatic-focus device. The VIDIAC (Vidicon Input to Automatic Computer) scans the image at high resolution and stores it temporarily on the Silicon Video Memory System (the heart of which is a new image-converter tube). The stored image can then be read directly into the computer's memory (through the Pattern Recognition Control System), or displayed on the grey-level TV monitor or the MACDAC interactive monitor. The MACDAC (Man Communication with Digital Automatic Computer) enables interactive communication with the computer by means of a "joy stick" cursor. The FIDAC (Film Input to Digital Automatic Computer) is a flying-spot scanner for scanning film transparencies at very high resolution; it is also used to make a print or film negative under the control of the computer, with high spatial and grey-level resolutions. The DRIDAC (Drum Input to Digital Automatic Computer) is a mechanical scanner of very great resolution in which a picture or X ray is wrapped around a drum and scanned by several photocells moving along a worm gear as the drum rotates.

All of the object scanning units (namely the VIDIAC, FIDAC, and DRIDAC) have systems that image and digitize the object picture; the digitized information can be sent to the core memory of the computer. Also, each of the scanned images can be stored on the Silicon Video Memory, in either video (analog) or digital form (i.e., a particular grey level or levels); many options can be selected by the user. From the Silicon Video Memory, the stored image can be displayed with grey levels on the TV monitor. The scanned images can also be stored on the MACDAC monitor (which is a storage tube). Then an interactive-graphics phase of a program can be performed in which the operator can present to the computer the x and y coordinates of a point, and

---

*This work was supported by U. S. Public Health Service grants GM-15192, GM-10797, HD-05361, and RR-05681 from the National Institutes of Health to the National Biomedical Research Foundation.

Fig. 1. National Biomedical Research Foundation Automatic Pattern Recognition and Computing Laboratory.

Fig. 2. Layout of the National Biomedical Research Foundation Pattern Recognition Laboratory and Computer Center.

NBR System 360/44

Fig. 3. National Biomedical Research Foundation 360/44 Configuration.

some coded information concerning the nature of the point. Finally, the computer itself can draw a picture on the MACDAC monitor, or on the TV monitor via the Silicon Video Memory System (or on a hard-copy printer via the Silicon Video Memory System). Thus the computer can present a MACDAC display, a TV monitor grey-level display, a hard-copy grey-level display, and a high-resolution grey-level display on film by means of the FIDAC.

The specialized pattern-recognition equipment of the laboratory includes:

(a) FIDAC (Film Input to Digital Automatic Computer) Mod II. The FIDAC is a high-speed high-resolution flying-spot scanner developed and built at the Foundation. It has a maximum capability of scanning a raster of 2000 x 2000, or 4,000,000 sample points within four seconds (i.e., at the rate of one microsecond per sample spot), with up to 16 grey levels per point. The number of sampled spots in the raster is adjustable, as is the number of grey levels. For instance, for the analysis of chromosomes by classical (nonbanding) methods, a nominal 700 x 500, or 350,000 sample-spot, raster is most efficacious. The analysis of fluorescent chromosomes is always accomplished from film, for the fluorescence fades as it is exposed to ultraviolet light. Thus a photomicrographic film is made in which the fluorescence from all the chromosomes can be recorded at the same time. (See Fig. 4.)

The FIDAC has two modes of operation, as follows: (1) A complete picture mode. In this mode the raster's size is nominally 700 x 500, that is, the picture is sampled at 350,000 spots, and each point can be sampled as black-and-white binary, i.e., one bit per spot, or up to 16 grey levels, i.e., four bits per spot. This raster size is used for the examination of biomedical photomicrographs which were obtained from an optical microscope, since this sampling rate exceeds the resolution of the optical microscope. (2) Control word mode. In this mode of operation a resolution of 2,048 spots across each of 1,365 rows (3 x 2 aspect ratio) with a maximum of 16 intensity grey levels can be obtained. This is accomplished by allowing the computer program to select all or any portion of the picture to be scanned with a corresponding sampling density such that the memory space allocated to the picture would not be overrun.

The specifications of FIDAC are as follows: A 35-mm still-frame film transport is used which can take up to 250 frames. The film frames can be advanced automatically under computer control. The logical circuitry utilizes integrated circuits soldered to printed circuit cards that are plugged into racks for ease of maintenance. The system is of our own design, completely synchronous, and can be clocked at any frequency up to 10 megahertz. The circuitry is geared to an easily adjustable master clock. Thus the clock speed can be increased or decreased, depending upon the characteristics of the computer to be used. The circuitry matching FIDAC to the computer consists of several replaceable cards. These FIDAC-computer interfacing cards are the only circuitry that would have to be changed in shifting from one computer system to another. The spot size is 1.5 mils at one foot-Lambert light output (measured on the face of the CRT). At the 1.5 mil spot size and at 2,000 spots per x-sweep of the raster, there is no overlap of spots because the x-scan raster is 3 inches wide. The rise time of the photo-multiplier is 2.3 nanoseconds, the video-amplifier bandpass is 10 MHz (megacycles), and the phosphor persistence is 100 nanoseconds. Thus one sample spot per microsecond clearly presents non-overlapping spots. Dynamic focusing is incorporated in FIDAC. This presents an almost constant spot size from the center to the sides of the CRT. The sampling rate used is 1 microsecond per spot. This means that the four-bit 16-grey-level analog-to-digital conversion takes place completely within this time period. One frame is scanned in about 0.35 seconds for complete picture mode (700 x 500 raster), and

Fig. 4. The Basic FIDAC System.

three seconds for a 2,000 x 1,365 raster. Jitter is related to the synchronization of the triggering of each successive line of the raster. The jitter of the FIDAC is less than 1 spot. Antipincushioning coils are included in both the x and y directions. By aspect ratio is meant the ratio of the distance between adjacent spots in the horizontal and vertical directions. It is usually desired to keep this at unity in order that no vertical or horizontal distortion occurs. On FIDAC the aspect-ratio is retained when the raster size is changed. The FIDAC can be used with 2 through 16 grey levels for each spot. Each grey level can be set within 0.1%. Shading refers to the correction of the grey-level variation across the area of the picture that arises from the CRT, photomultiplier, and optical systems. For the FIDAC, the shading generator can be adjusted so that for a blank frame an almost uniform response is obtained across the whole frame. If successive scans are made on the same picture frame, then the starting point, or very first spot, on such successive frames will be at the same place in the picture frame to within one spot diameter. The raster size can vary from line to line in length of line or in the rate of scanning within the line. The stability of such possible errors is within one spot diameter. The cutoff voltage of each level detector is stable within .01%. The drift in the output of the video amplifier has been stabilized to maintain a constant output for a given input signal. A feedback loop adjusts the average integrated CRT-spot intensity of the frame to a constant value.

(b) VIDIAN (Vidicon Input to Automatic Computer). The VIDIAN device was also designed and constructed by our electronic laboratory either for scanning through a microscope or for scanning large pictures or X rays. It consists of a high resolution vidicon system that scans at standard TV rates (i.e., 1/30 sec/frame), coupled to a Silicon Video Memory System. The standard vidicon scan rate is too fast for most analog-to-digital conversion systems and for data transmission to a digital computer. Hence the Silicon Video Memory System acts as an interface between the vidicon and the computer. The capabilities of the VIDIAN are as follows. The image scanned by the vidicon can be displayed on a TV monitor for viewing and for optimizing such parameters as focus, intensity, contrast, etc. The image scanned by the vidicon can be written onto the Silicon Video Memory System. The electronic image as stored in this silicon memory can in turn be internally scanned and displayed on either the TV monitor or on the storage tube of the MACDAC. A zoom control on the silicon memory and a bias control on the console of the MACDAC can be used to display any portion of the electronic image or to read into the computer any portion of the image. This zoom feature is also under computer control. The standard vidicon scan rate of 1/30 sec. per frame is equivalent to about a 30 megahertz spot sampling rate; this fast scan is the rate from the vidicon to the silicon memory tube and from the silicon memory to the TV monitor. The equivalent spot sampling rate from the silicon memory to the computer or to the MACDAC, or of writing onto the silicon memory from the computer, called the slow scan, is a maximum of a megahertz, but frequently 1/3 megahertz is used. Switching the flow of information between the vidicon, TV monitor, MACDAC, and computer, together with the appropriate scan rates, is under both manual and computer control. (See Fig. 5.)

(c) SPIDAC (Specimen Input to Digital Automatic Computer). The SPIDAC is an instrument for scanning directly through a microscope, digitizing the resulting grey-level pattern, and placing the digitized image directly into the computer's memory. The system is able to automatically manipulate the microscope slide in both the x and y directions, record the coordinates of objects of interest, automatically focus the image instantaneously, and monitor results, all under computer control. Eliminating the photographic

Fig. 5. Block Diagram of VIDIAC System.

step has the advantage of saving time and effort. Also, in a number of instances it is possible that such direct digitization may bring more sensitivity to grey-level discrimination and can eliminate certain possible grey-level distortions which can occur when using film as an intermediate step. (See Fig. 6.)

One of the main functions of the SPIDAC is to locate specimens of interest on the glass slide by automatic means, provided that an object of interest can be specified utilizing pictorial parameters. The SPIDAC system performs its operation in two phases. During the first phase a coarse scan (40x objective) is made to determine where the objects of interest are located. During the second phase a final and more accurate determination (at 100x objective) is made of whether or not the object located in the first stage is of sufficient interest to be automatically analyzed, and if so then its image is scanned by the VIDIAC and digitized and read into the computer. The pattern-recognition programs in the computer are utilized to compute the desired quantitative information from the pictorial data. The 40x and 100x objectives are parfocal, and a filter is introduced when the 100x is in place so that no adjustment of the vidicon is necessary when going from 40x to 100x objective.

In particular, SPIDAC scans a 1 cm x 1 cm field on the glass slide. At 40x objective, the frame size is $200\mu \times 260\mu$, giving about 40 strips of 50 frames each (or 2,000 frames) over the 1 cm² area field. Each of the x and y stepping motors is run at a rate of 650 steps/second, with 200 steps/revolution, moving the stage 500μ /revolution. This amounts to 2.50μ /step, meaning that we can locate any point in the field to the nearest 1.25μ . The rate of movement is $(650 \times 500) \div 200 = 1,625\mu$ /second, or $10,000 \div 1,625$ or about 6 seconds/strip, or finally $6 \times 40 = 240$ seconds for the 40 strips of the field. To move from strip to strip will take a total of 6 seconds across the field, so the total scan time limitation due to the motors themselves is 246 seconds or 4 minutes and 6 seconds. In addition, each field is examined (in the first phase as described above) for a chromosome spread, and this takes $(1/30)$ second per frame, or $2,000 \times (1/30) = 66.6$ seconds or 1 minute and 6.6 seconds. Thus the total time that SPIDAC takes to search a 1 cm x 1 cm field area for chromosome spreads is 5 minutes and 12.6 seconds.

(d) MACDAC (MAN Communication with Digital Automatic Computer). The MACDAC is an interactive-graphics device designed and constructed by us for communicating with the computer. Automatic chromosome analysis involves a limited but essential manual "editing" step which is accomplished by means of the MACDAC. After the computer console is properly set up, the control of the complete chromosome analysis is carried out by the operator at the MACDAC console. The expanded MACDAC will include a hard-copy grey level picture printer. Two monitors are available to the MACDAC. The first is a Tektronix storage tube capable of only black and white (actually green) display; the second is a TV monitor capable of grey level display. The interaction is carried out by means of a joy stick controlled cursor. For any position of the cursor, its 10-bit x and 10-bit y coordinates can be read into the computer, together with a four-bit code set by the operator, on a priority interrupt basis. This four-bit code identifies for the computer the meaning of the x and y coordinates. The interrupt capability allows the relatively slow interactive editing process to be carried out at the same time as the computer is executing an analysis program, on an overlapping basis. (See Fig. 7.)

Fig. 6. Block Diagram of SPIDAC system (including the associated parts of MACDAC and COMPUTER hardware and software).

Fig. 7. (a) MACDAC Console: At left is the display tube; at right, the control panel. The lower left (main) joystick controls the cursor; the upper right (bias) joystick controls the zoom feature. The line of five switches below the bias joystick controls the four-bit code which is read in with the coordinates when the "Read Spot" button is pushed. Other, nonstandard codes can be set by means of the five switches below the main joystick.

(b) Some Details of the MACDAC Design.

The MACDAC has two joy sticks, one for normal interaction, and the second for zoom control. The zoom control enables any area of the display to be enlarged so that detail can be more easily observed. The MACDAC console includes switches to control the scan of the FIDAC or VIDAC, to tell the computer that the interactive procedure has been completed, to direct the computer to display a picture on the monitors, to tell the computer to enable the use of the joy sticks, and so forth. Either the analog video signal or the digitized signal can be displayed on the MACDAC. For the digital signal, any particular grey level can be seen, or any grey level above a certain cutoff level can be displayed, and so forth.

(e) DRIDAC (DRum Input to Digital Automatic Computer). Many pictures of biomedical importance are produced in relatively large sizes, both as transparencies and black and white prints. Some examples are large chest X-ray plates, X rays of large bones, X-ray crystallographic plates, angiograms, graphs such as electroencephalographs and electrocardiographs, and many others. The DRIDAC device was designed and built at the National Biomedical Research Foundation to allow such pictures to be directly read in to a digital computer. (See Fig. 8.)

DRIDAC is a high-speed, high-resolution, on-line drum scanner which can scan photographs (X ray, transparency, or print) of any size up to sixteen inches by twelve inches of useful area. The resolution may be varied to as fine as 0.004 inches. This allows it to sample 4,000 x 3,000 spots, for a total of 12 million spots, from a large picture. Each spot can be digitized as required into one bit (black and white) or up to six bits (sixty-four grey levels per spot). An automatic background detection and compensation feature when activated affords a much greater dynamic range and in effect increases the number of grey levels. Another innovation in our drum scanner is the utilization of two read stations (see figures) thereby reducing the total scan time to one-half the time required by a single read station.

In consideration of the large quantity of data which can be obtained from a single picture, we have designed the DRIDAC device to read the pictorial data into the core memory of our IBM 360/44 computer at a maximum rate of 252,000 bits per second and simultaneously transfer the accumulating data from the core into the IBM 2311 disk drive pack (see Fig. 9). Since a 2311 disk pack will store approximately 54 million bits, this requires a slight reduction in grey-scale resolution if we are to read in a full 4,000 x 3,000 spot picture.

The DRIDAC's drum speed is 420 rpm and its sampling rate is approximately 35 microseconds per spot. Thus to read in an entire picture requires approximately five minutes. The DRIDAC's high speed in getting a large quantity of data onto the disk saves valuable computer time. Once the data is on the disk, the program can select any desired portion of the picture at disk speed. It would be inappropriate to process data directly from the drum scanner, since backtracking or moving to distant spots under the control of the computer is slow and would therefore waste computer time.

Of no less importance than the hardware (or computer instrumentation) is the associated software (or computer programming) systems that we have developed and are available for use in this work. Since the programming systems or languages are extremely extensive and complex, we can only give the "flavor" of the languages in the following discussions. These software systems include:

The specialized pattern-recognition software of the laboratory includes:

DRIDAC TIMING

BLACK MARKS = LOGIC 1
WHITE BAND = LOGIC 0

Fig. 8. DRIDAC System.

PICTURE FORMAT WITH 2 HEADS

Fig. 9. Illustration of System for Interleaving the Two Halves of the Picture onto the Disc.

(a) FIDACSYS (FIDAC system). This system inputs the images into the computer from the scanning devices, and accomplishes overall picture manipulation and enhancement. The list of options and subroutines is too lengthy to give here, but we will illustrate a typical set of options available for reading from FIDAC or VIDAC into the computer.

One part of the system is a general utility program for handling FIDAC pictures conveniently. It may be used to read FIDAC, VIDAC, or DRIDAC, store pictures on tape and read pictures from tape, and allows several options such as specification of bits per spot from 1 to 6, and limited printing windows designated by using the joy stick. Up to fifty character sets are stored on disk "FIDACD" and any one may be called upon to print a picture. These character sets may be altered and the complete set may be listed for examination at any time. The program can be interactive with the computer console and MACDAC, and in this mode responds to one "command" at a time, and upon completion types the word "COMMAND?" In general, each command consists of two letters and two numbers.

A typical command is CB, change bits: this allows change of spatial density read from FIDAC and number of bits per spot. Spatial density may be any integer from 1 through 15. Bits per spot may be 1, 2, 3, 4, or 6. With this information, the computer determines whether an entire picture of this size will fit in memory. If it will not fit, the computer responds by requesting a window size to be pointed out on the MACDAC scope using the joy stick. This window is designated by indicating the width desired, and the computer responds by indicating the height allowed for that width by drawing the picture limits on the scope. This procedure may be repeated until the operator is satisfied with the area he will get. When satisfied, he pushes the END button, and the window is stored in memory. Often a picture size of exact dimensions is necessary, when reading a picture from tape, which could not be easily obtained by pointing on the scope. This may be specified as follows: when the computer has requested the joy stick coordinates for the top corners desired, push the INTERRUPT button. The computer will now ask for the exact number of spots per row. The number of lines is calculated and is not under the operator's control. The END button must be pushed to accept the results of this command, unless a non-zero number is typed after spots per row is typed.

Other commands are: CCnn, change characters; CS, change size; CW, change window; PC, print characters; WF, write on film in FIDAC (i.e., make a grey level negative on the FIDAC used as a "reverse" scanner); RFnn, read FIDAC; RTnnKK, read tape; SP, spectrum point; PPnnKK, print picture on high speed printer with a designated character set; WSnn, write on the MACDAC display unit; WTnnKK, write tape, and so forth.

For reading very large pictures (i.e., raster sizes of 2,000 x 2,000 spots) a special subprogram is required as follows. The subprogram "FIDPIC" reads a FIDAC picture into memory from the FIDAC, but retains only a specified window in memory. To do this, the program constantly monitors the address into which the picture is being read; when an undesired portion is being read, this address is altered to reference a "discard" area and that portion of picture is thus destroyed. Using this technique, the picture can be read, portion after portion, on successive scans, each scanned region stored on disc and thereby assembled as a complete picture which would have been too large to be stored in memory in its entirety.

Programs "FIDIND" and "REDUCE" are used in conjunction to read a 4 bit/spot picture from disc and convert to 2 bits/spot in memory with the resultant loss in grey levels, but gain in picture size capable of being stored.

(b) SYNTAXSYS and BUGSYS Pattern Recognition Languages. These two systems present a large variety of basic pattern-recognition capabilities, including syntax-directed pattern recognition, feature-recognition methods, boundary-characterization techniques, etc. We have published extensively in the literature on these languages.

The SYNTAXSYS language uses an approach to pattern recognition based on some of R. S. Ledley's early original research into picture grammars. Consider a characterization of the boundary of an object in terms of five types of curves: a clockwise curve, type A; a relatively straight line, type B; a counterclockwise curve, type C; a notch, type D; and a wide clockwise curve, type E. Syntactical definitions of the different kinds of chromosomes can be made in such terms as these. A boundary is first characterized as a list of such curve types, and then the syntactical definitions are used to "build up" derived parts of a chromosome from combinations of these curve types. For example, the syntax for a submedium chromosome and a telocentric chromosome is shown in Table 1. The recursive definition

$$\langle arm \rangle :: = B \langle arm \rangle \mid \langle arm \rangle B \mid A$$

means that the generic concept *arm* is defined as being a B type followed by an arm, or an arm followed by a B type or an A type. In this notation, the angular brackets < > indicate that a generic name is enclosed, the ":: =" means "is defined as being," and the "|" means "or." The definition is recursive in that it is used repeatedly. For example, consider our definitions applied to the figure ABC as follows:

where we first (I) identify the A type as an *arm*, and then build up the arm by repeated application of the definition to include (II) the B type on the left and (III) the B type on the right. Further consideration of these concepts can be found in the references. (See Fig. 10.)

The process is, of course, systematic, and is carried out by a special program in the SYNTAXSYS system called the MOBILIZER [which is directly analogous to the "translator" of automatic-programming-language translation]. The mobilizer operates on a parts list for a particular object, and by using the generic syntactical description of various kinds of chromosomes "recognizes" the object as a chromosome or not a chromosome, and if a chromosome is recognized, as a particular kind of chromosome. Actually the mobilizer works with numbers: as shown in Table 1 the syntax can as well be written in terms of level numbers, where the numbers are assigned to the left-hand entities in the order in which they appear in the syntax. The basic parts, namely A, B, C, D, and E, are always considered to have number values less than any generic or "derived" part.

Fig. 10. (a) The basic types of curves used for the illustration in the text.
 (b) Illustration of submedian and telocentric chromosomes.
 (c) Short example used in text.
 (d) Example of syntactical analysis or the build-up process followed in chromosome recognition for the chromosome shown in (b). Here *am* stands for <arm>, *rp* for <right part>, *ap* for <arm pair>, and *s* for <side>.

Table 1. Sample syntax for submedian and telocentric chromosomes.

$\langle \text{arm} \rangle :: = B \langle \text{arm} \rangle \mid \langle \text{arm} \rangle B A$	70 :: = B,70 70,B A
$\langle \text{side} \rangle :: = B \langle \text{side} \rangle \mid \langle \text{side} \rangle B B D$	71 :: = B,71 71,B B D
$\langle \text{bottom} \rangle :: = B \langle \text{bottom} \rangle \mid \langle \text{bottom} \rangle B E$	72 :: = B,72 72,B E
$\langle \text{right part} \rangle :: = C \langle \text{arm} \rangle$	73 :: = C,70
$\langle \text{left part} \rangle :: = \langle \text{arm} \rangle C$	74 :: = 70,C
$\langle \text{arm pair} \rangle :: = \langle \text{side} \rangle \langle \text{arm pair} \rangle \mid \langle \text{arm pair} \rangle \langle \text{side} \rangle$	75 :: = 71,75 75,71
$\mid \langle \text{arm} \rangle \langle \text{right part} \rangle \mid \langle \text{left part} \rangle \langle \text{arm} \rangle$	70,73 74,70
$\langle \text{submedian chromosome} \rangle :: = \langle \text{arm pair} \rangle \langle \text{arm pair} \rangle$	76 :: = 75,75
$\langle \text{telocentric chromosome} \rangle :: = \langle \text{bottom} \rangle \langle \text{arm pair} \rangle$	77 :: = 72,75

The BUGSYS language is a picture-processing and -measuring programming language for the analysis of the picture in the computer's memory. The main concept of the system is the use of a collection of programmable pointers, which are visualized as a family of "bugs." A bug can be "initiated," or "placed," and once initiated a bug can be "moved." In addition, a bug can "change" the grey-level value of the spot on which it is located, and it can lay down a so-called "stick" across a thick line in the picture as an aid to locating the middle of the line. Also, two bugs together can "probe" along the direction of the line between them or along a direction perpendicular to the line between them. These probes can sense an extension of an object or the width of an object, and so forth. "Globs" of objects can be assessed by laying down squares and determining the percentage of the area of the square intersected with the object. The system is composed of many such statements as will be illustrated below. (See Fig. 11.)

For each bug initiated by means of the macro PLACE, there is associated a list which gives the x and y coordinates of the current position of that bug, the actual core location, the spot position within this location, and the grey-level value of the current position of the bug. For example, if the bug named "ZIPPY" is in the position $x = 16$, $y = 8$, then the list for ZIPPY might contain the following.

Address	Contents	Comment
ZIPPY(1)	16	x
ZIPPY(2)	8	y
ZIPPY(3)	6351	6000 (first location of picture) +350 [50 words per line $\times (y-1)$] +1 (x' divided by 10 spots per word and truncated to an integer)
ZIPPY(4)	6	remainder of x divided by 10
ZIPPY(5)	4	gray-level value or contents of spot

As a bug is moved about the picture by a program, the list of the bug is kept current. In fact, in essence, this list is the bug.

We will now proceed to describe some of the statements of the BUGSYS language. The PLACE statement initiates, or sets up, a bug by assigning a name and initial coordinates to it. Thus we have

Fig. 11. Illustration of the Application of BUGSYS to Determining Successive y Coordinates of a Curve (taken from a Schlieren photograph). The dots illustrate the path of the bugs taken in making the successive Δy measurements.

<PLACE statement> :: = PLACE <bug name>, <x coordinate>, <y coordinate>

where the bug name is a FORTRAN label and the x and y coordinates are either unsigned integers or integer variables.

The BUGS statement allocates five storage locations to each bug named; its form is

<BUGS statement> :: = BUGS (<bug-name list>)

where the bug-name list is a string of bug names separated by commas.

A bug can be moved a specified distance (i.e., number of spots in the picture) in either the x or y direction by the following statement.

<MOVE statement> :: = MOVE <bug name>, <direction>, <distance>

where the direction is given by the literals LEFT, RIGHT, UP, or DOWN, and the distance is an unsigned integer or integer variable. The statement "MOVE ZIPPY, RIGHT, 15" moves the bug named ZIPPY to a new location having the same y coordinate but a new x coordinate 15 spots to the right of the present x coordinate. Each time a bug is moved, of course, the list corresponding to the bug name is adjusted for the new values.

Many of the statements involve multiple-way branches, for provision must be made to be sure that the bug will not be moved out of the picture. For instance, in the MOVE statement if the bug would be moved out of the picture, then it is not moved at all, and the next sequential instruction is taken as a next executed instruction. Otherwise (i.e., if the bug will still be in the picture after the move) the bug is moved and the next sequential instruction is skipped. The BUGSYS language also includes provisions for bounding an object. This is the same subprogram as the bounding routine which appears in the SYNTAXSYS system, except that in BUGSYS it can be utilized in a macro called BOUND.

(c) MACDACSYS (MACDAC system). This programming system implements the interactive capabilities of the MACDAC unit, including interrupt features for accepting information from, and displaying pictorial and alphanumeric information on, the MACDAC unit. Typical examples of operator interactive capabilities provided by the MACDACSYS system are as follows. Utilizing the joy stick in either regular or zoom mode, the operator can connect two points, erase an object, slice an object into two parts, and draw a "fence" or line around a collection of objects that are to be erased from the picture in the computer's memory. Also the "end of job," the "end of codes," and an "erase last code put in" commands are available. In many interactive-operator procedures, the objects displayed on the MACDAC display are numbered. The operator can point to one of the numbers to direct the computer to process in some way the associated object. To do this the joy stick is positioned with the cursor near the number and the "read spot" button is pressed. If the computer's bell rings (the IBM 360/44 has a programmable bell) then this tells the operator that the cursor is not close enough to the number. This method of having the computer signal the operator by means of the bell, or a series of coded bell rings, is used extensively in our interactive system.

The MACDACSYS system is designed for capability of running in a time sharing or overlapping mode. In this mode, the picture being worked on has been placed on the MACDAC display directly from the scanners, and not from the computer. Thus the computer can be carrying out an entirely different computation with no picture at all, or a different picture, in its memory. The editing or interactive procedures carried out by the operator are stored in the computer's memory, the MACDACSYS interrupting the computer's computations from time to time as the operator is working. Then, at a later time when the operator actually scans the picture into the computer's memory, the directions given previously by the operator for editing the picture are now carried out.

(d) SPIDACSYS (SPIDAC system). This system automatically directs the motion of the SPIDAC microscope stage, detects good chromosome spreads, and records the coordinates (to the nearest 1.25μ) of the center of each such good chromosome spread. It also relocates the microscope stage at such coordinates, and directs the vidicon scan into the computer.

The SPIDACSYS software package consists of two parts. The first part is pertinent to the covering, detection, and locating centers of the objects of interest (e.g., chromosome spreads) during the search mode. The second part controls the stepping motors and moves the microscope stage such that these objects are centered automatically for detailed analysis during the analysis mode.

The complete operational procedure is as follows: The operator or the technician first puts the glass slide onto the microscope stage, selects the origin (usually left-upper corner of the cover slip), and turns to the low power object lens. At the same time, he loads the software package into the computer. When everything is ready, he pushes the start button and waits. Meanwhile, the hardware searches the successive fields of a strip, and then goes to the next strip, etc. until the forty strips have been searched field by field. In between searches the software program finds covers and determines centers of the objects of interest (e.g., chromosome spreads) in each field, and this information is saved for the analysis mode. An optional visual display is also included in the software program. This visual display puts all the coverings and domains of chromosome spreads on the MACDAC scope (or TV screen) for each field. (See Fig. 12.)

Next the operator or computer switches the objective lens from low power to high power. He then pushes the start button for the analysis mode. The computer program sends a signal to the hardware to move the microscope stage so that each of the objects found is successively centered under the objective lens. The hardware sends the detailed scan of each such object one by one onto a TV screen. The technician judges from the TV screen the acceptability of the object for analysis. If it is a bad one, he may reject it and the next object will be shown. If it is a good one, he may push the analysis button. This enables the fine scan of the object to be made and the object is analyzed by the computer. Then the next available object will be displayed. This procedure continues until all the objects have been processed. The software program then moves the microscope stage back to its original position and terminates the procedure.

(e) DOCS (Display Of Chromosome Statistics). The DOCS programming system enables on-line computer-console interaction with the disk memories of the computer for the evaluation and display of large masses of data in

(a)

(b)

Fig. 12. Application of SPIDAC to Detecting Chromosome Spreads.

- (a) Chromosome spread as seen on MACDAC display tube.
- (b) Horizontal lines illustrate the recognition of the portion of a chromosome spread.
- (c) Two chromosome spreads in a field.
- (d) Result of computer locating each of the chromosome spreads.

a file. For instance, analyses of variance, t-tests, and other statistical tests can be executed on data selected from the file. The DOCS system guides the user in a step-by-step dialogue, so that all of the detailed specifications required for the desired statistical analysis or data display can be obtained.

(f) REMOTE. This programming system enables the remote user to be serviced by the computer.

REFERENCES

1. R. S. Ledley, Programming and Utilizing Digital Computers, New York: McGraw-Hill, ch. 8, 1962.
2. R. S. Ledley, M. Legator, and J. B. Wilson, "Automatic Determination of Mitotic Index," in Pictorial Pattern Recognition, G. C. Cheng et. al., Eds., Washington, D. C.: Thompson Book Co., 1968.
3. R. S. Ledley, "High-Speed Automatic Analysis of Biomedical Pictures," Science, vol. 146, no. 3641, pp. 216-223, October 9, 1964.
4. R. S. Ledley and F. H. Ruddle, "Chromosome Analysis by Computer," Scientific American, vol. 214, pp. 40-46, April 1966.
5. F. Ruddle, S. Smith, R. S. Ledley, and M. Belson, "Replication-Precision Study of Manual and Automatic Chromosome Analysis," Ann. N. Y. Acad. Sci., vol. 157, art. 1, pp. 400-423, March 31, 1969.
6. T. Golab, R. S. Ledley, and L. S. Rotolo, "FIDAC - Film Input to Digital Automatic Computer," Pattern Recognition, vol. 3, pp. 123-156, 1971.

THE CAMAC STANDARD AND ITS USE IN
COMPUTERIZED MEASURING SYSTEMS

K. Čuljat, R. Esteverena, T. Douglass

ORTEC, Inc.
Oak Ridge, Tennessee

ABSTRACT

The CAMAC standard provides standardized method for transmitting data and control signals between digital computer and the instrumentation modules. Mechanical features, dataway, branch highway, and software aspects of this modular instrumentation system for data handling are briefly described in this paper. Advantages of the standardized and modular system are pointed out and an example of the implementation is shown as well.

THE CAMAC STANDARD AND ITS USE IN COMPUTERIZED MEASURING SYSTEMS

K. Čuljat, R. Esteverena, T. Douglass
ORTEC, Inc.
Oak Ridge, Tennessee

1. INTRODUCTION

Recently, a major step in interfacing data acquisition and controlled instruments with a computer or digital controller has been taken with the introduction of CAMAC. This acronym stands for "Computer Activated Measurement and Control," a standard developed under the auspices of the European Standards Organization for Nuclear Energy, a Euratom organization with representatives from 12 countries. Although developed under ESONE, the U.S. Atomic Energy Commission recognized its value, and its NIM Committee has endorsed the CAMAC system as "a Dataway system complementary to NIM." CAMAC and NIM are complementary and compatible. The CAMAC system is concerned primarily with the transfer of information in the form of digital signals; it provides a digital interface between controlled instruments and a computer. The NIM system is used primarily in data handling of analog signals.

There are various reasons for standardization. Probably the most important is system compatibility. A user should be able to put together a system composed of units produced by many manufacturers without the need for special interfacing devices, which he would probably have to build himself. Solving the problems posed by the need to integrate two instruments with different logic, power supply, and connector requirements represents a large waste of time. Now the user can choose the best instrument and system for his needs from various suppliers, knowing that it will be completely compatible with the rest of his system. Other benefits to the user include the assurance of quality and the simplicity of instrument specification.

The user isn't the only one who derives benefits from a standard instrument system. The manufacturer also gains some advantages. Standardization can actually save money in the long run. It enables the engineer to go about the business of designing new functions without having to worry about packaging, mechanical considerations, and power supply design. In addition, standard packaging offers possible economies in large-scale manufacturing that benefit both the user and the manufacturer.

Both CAMAC and NIM are standard modular instrumentation systems. The use of modules in instrument systems has also proved to be useful. To the user the simplicity of expandability is quite apparent. He starts with a system that meets his requirements of the moment and later expands the system as his requirements change. Similarly, modules make it practical to establish an instrument bank from which various individuals can select equipment. Another important feature is the short down-time of a system caused by instrument failure. One simply replaces the module that has failed and proceeds with data-taking.

As is obvious, of course, no standard is optimal for all purposes. For a given problem it is possible to design a data acquisition system that more

closely satisfies a particular need. On the other hand, the long range of benefits of standardization, such as interchangeability, reusability, and compatibility of commercially available instruments make the advantages of standardization obvious.

Before going into detail it is very illustrative to view at Fig. 1 which is a photograph of some instrumentation at the Lawrence Radiation Laboratory in Berkeley in about 1963. It is out of question that a vast amount of digital data can effectively be handled in this way. Some sort of Dataway system was inevitable. CAMAC thus has been developed and is already widely accepted in Europe and in the United States.

CAMAC, like NIM, was originated by requirements for nuclear instrument systems. The use of CAMAC in nuclear applications is already widespread, but both NIM and CAMAC as standard modular instrumentation systems are by no means limited to Nuclear applications. Some of the nonnuclear applications include a number of different measuring and experimental systems, for example, in x-ray, acoustical, meteorological, and biomedical research areas.

2. DEFINITIONS AND AN OVERVIEW OF THE CAMAC SYSTEM

Since CAMAC is a modular instrumentation system, it is comprised of modules. A module, Fig 2, is normally a printed circuit board with integrated circuits mutually interconnected, a front panel with some controls, connectors and/or indicators, and provision for connection with the edge connector on the rear end of the board. Each module typically has two parts. Part A is a digital circuit such as a scaler, parallel input register, interrupt handler, ADC interface, etc. This A part is the one that performs the function desired and is the real reason for existence of the module. Part B is the part that "speaks CAMAC," i.e., facilitates the conversation between part A and the "brain of the system."

In the CAMAC system there must be means of physically housing the modules and connecting them electrically to the whole system. This task is fulfilled by a crate, Fig. 3. The crate is a box with 25 slots to house up to 25 single-width modules. The rear of the crate contains a multilayer board with a Dataway. The Dataway (DW) is the wiring that interconnects 25 card edge connectors, through which the module can have access to the Dataway's data, control, and power lines. The power supply is normally a separate entity, rack-mountable, below the crate.

Each crate has, in addition to the modules already described, one double-width control module, the Crate Controller (CC). It controls the flow of signals in the crate and provides common interfacing between the modules and the system controller, which might be either a computer or any special-purpose control device.

In large systems more than one crate is needed. In these multicrate systems, crates are mutually interconnected by means of a Branch Highway (BH), Fig. 4. Crates are connected through their CC's to the BH in a daisy-chained way. The BH is terminated on one side by the terminator, and connected on the other side to the Branch Driver (BD). The BD controls the flow of signals in the branch (complex of BH and crates with modules and CC's) and provides the interface with the computer.

In the multicrate system just described, CC's could be standardized and their implementation would not be computer-dependent. Such a CC is called a Type A CC. The only computer-dependent instrument, in terms of design, is the BD, which must be changed if the computer is replaced by another one.

3. DATAWAY STRUCTURE AND THE FUNCTIONS OF THE DATAWAY SIGNALS*

Each slot in the crate can house one single-width module and is called a station. Stations are numbered from left to right (front view) by numbers 1

*Described and specified in EUR 4100e document.

Fig. 1. Some Instrumentation at the LRL at Berkeley in 1963.
(Courtesy of National Bureau of Standards)

Fig. 2. CAMAC Module

Fig. 3. CAMAC Crate

Fig. 4. Multicrate System

through 25, Fig. 5. Stations 1 through 23 are intended to be occupied by "normal" modules. A normal module is any module except the CC.

Modules are connected to the Dataway through card edge connectors and can have single, double, triple, or any multiple width. Stations 24 and 25 are occupied by the double-width CC, which is connected to the DW by the two connectors. Most of the DW lines are bussed, i.e., the same signal is available at the same pin on all connectors 1-24, Fig. 6. The other lines are radially oriented, fanning out from the connector at the station 25, each line being associated with one station from 1 to 24.

All DW signals are carried by associated DW lines, which is why both signals and lines have the same name. In the following discussion reference is made to the DW signals whenever the intent is to emphasize the function or to the lines when geometry is considered. One-to-one correspondence between DW signals and lines should be kept in mind.

Addressing

The first group of radially oriented lines carries signals designated N, indicating the station number. Each station is addressed by its associated individual line from station 25 of the CC. The connector at station 25 has 24 N pins, N1 through N24, and connectors at the other stations have only 1 N pin; therefore the module is addressed by the number of the station it is plugged into. The module address is not coded, and so it is possible to address any subset of modules simultaneously.

To permit distinction between different parts of the addressed module, each module can have up to 16 subaddresses A(0), A(1),...A(15). Subaddress information is carried by four bussed lines, A1, A2, A4, A8, in binary coded form and is available at each module connector. Proper subaddress within the addressed module is selected by decoding the information from the A lines. Decoding logic is physically placed in the B part of the module.

Functions

Modules should perform some function F, which is specified in a binary coded form on the five bussed lines F1, F2, F4, F8, and F16 available to all modules. Function information, determined by the CC, is decoded by the decoder in the B part of the module. There are 32 possible different functions F(0), F(1),...F(31), such as read, write, test, and control functions.

Most DW operations are commands issued by the CC. Functions just described are addressed commands since function F is always associated with the module address N and subaddress A. N and A together select the address of a specific register or piece of logic within a module, and F specifies the function to be performed by the selected subaddress. The major category of CAMAC commands are these NAF commands and are directly related to CAMAC existence.

Data Signals

There are two sets of bussed data lines: read bus and write bus. The read bus contains 24 lines, R1 through R24, which enable parallel transfer of a 24-bit data word (or smaller) from module to the CC. Write bus is another set of 24 lines W1 to W24, which permit transfer of a 24-bit data word (or smaller) from CC to any module in the crate.

In either read or write transfer, only modules addressed by the NAF command will be involved even though all modules have access to both busses. In other words, data are gated by proper NAF command from the module to the read bus or from write bus to the module. The necessary gating network is built into the B part of every module. There is no standard word size in CAMAC: any word size is legal, with the only limitation being that the maximum word size is 24 bits.

Fig. 5. Crate with Numbered Stations

Fig. 6. Dataway Bussed and Radially Oriented Lines

Timing

The transfer of any information or the performance of any function occurs during strobe pulses provided by the CC. The CC contains a timing generator that generates two strobe pulses, S1 and S2. One full DW operation is performed in one DW cycle. The DW cycle is the time period from the moment when NAF command is issued, during strobes S1 and S2 until termination when NAF is removed from DW, Fig. 7. S1 is the strobe for timing the first phase of the DW operation, S2 the strobe for the second one. The duration of the DW cycle must not be shorter than 1 microsecond, and longer cycles are permissible. Real duration is fixed by the CC, and each module must be designed to have "access time" short enough to be compatible. Operations on the DW are not handshake type. Strobe pulses are carried by bussed lines S1 and S2, and the proper gating logic should again be included in the B part of the module.

Nonaddressed Commands

In addition to addressed NAF commands, there are also the unaddressed commands, initialize (Z), inhibit (I), and clear (C), which are bussed to all stations. Initialize is used to force the system into a defined state when power is first turned on or whenever a complete restart is required; clear is normally used to clear data registers; and inhibit can be used to prevent some actions. Considerable freedom is left to the module designer as to how to use these signals.

Status Lines

There are other bussed lines that carry status information. These are busy (B), response (Q), and look-at-me (L). B indicates to the CC that DW operation is in progress and that another DW operation cannot be started yet. Q is a bussed line that indicates whether the module has performed properly or not.

Interrupt Feature

It is well known that a computer interrupt feature enables devices to demand computer attention. How is that implemented in CAMAC? There is a set of look-at-me (L) lines fanning in radially to the CC from all stations in exactly the same way that N lines fan out from the CC. So the CC connector at station 25 has 24 L pins (L1 through L24) and connectors at the other stations have only one L pin. When a module is ready to communicate with the computer and requires the computer to undertake some action, it says "look-at me", that is, it pulls its own L line, which will cause an interrupt. The computer should be programmed to respond properly to that interrupt signal.

By simple calculation it could be easily verified that there are 23 times 16 theoretically possible interrupt sources in one crate. In reality that number will never be reached. The problem of identifying an interrupt source and servicing it is neither simple nor standardized. It is pointed out that all 24 lines are usually OR'ed in the CC to generate a unique interrupt request. Having access to all 24 L lines, the CC knows which modules have requested interrupt. The situation is complicated by necessary priority arbitration if several modules issue a request simultaneously and by determining from the source subaddress whether more possible "subsources" exist within each module. To solve this problem, combined hardware and software methods might be used. In type A CC, described in EUR 4600e, 24 L lines are OR'ed with 24 R lines so that the "L pattern" could be read by the computer. It is also possible to identify an interrupt subsorce by reading the interrupt flags within the module (B part again).

Fig. 7. Dataway Timing Cycle

4. BRANCH HIGHWAY AND SYSTEM OPERATION*

In the large multicrate system it is necessary to carry DW information from individual crates to the computer. This is done by means of the branch highway (BH), which connects crates through their CC's to the branch driver (BD), which, in turn, is interfaced to the computer, Fig. 4. The BH standard implies a standardized type A crate controller (CCA). Unlike the DW, the BH is a multiwire cable with common bidirectional read-write lines. While DW operations are synchronized, BH operations are asynchronous, sometimes referred to as the handshake type.

The type A crate controller transforms some signals between the DW and the BH and provides almost direct transfer of the others, after possibly one level of gating. All branch signals are preceeded by the upper case letter B.

Crates are addressed within the BH similarly to the way that modules are addressed within the DW. The BH can have up to 7 crates and there are 7 BCR lines for the crate addressing, BCR1 through BCR7. All crates have access to all BCR lines, the particular BCR line being selected by the 7-position switch on the CC.

After a particular crate is selected by the associated BCR line from BD, all other BH signals are gated by this BCR signal in the CCA. DW operation in the addressed crate is initiated by the timing signal BTA, which represents the first action in the handshake communication between the BD and the CC. Using BH signals BN, BA, and BF, a NAF command is issued by the CC. Undirectional R and W DW lines are connected to the bidirectional BRW (branch read-write) BH lines, depending on whether read or write is to be performed. As a consequence of the BTA timing pulse, S1 and S2 pulses are issued by the CC timing generator with a proper delay. Upon completion of the DW operation, the CC responds to the BD by sending a BTB timing pulse. This pulse indicates to the BD that a DW cycle is complete and that the BD can start another operation.

While the signals on BA and BF lines are fed to the A and F lines without modification, BN lines for station addressing are modified in the CC. There are five BN lines in the BH which carry binary-coded station-number information. They are fed to the decoder in the CC, which then generates the already described Dataway N signals.

The BH is terminated on one end by a terminator as a normal transmission line and on the other end by BD, Fig. 4.

The BD is a more sophisticated instrument than the CC since it includes the interface to the computer as well as logic to control operations in the whole branch. Generally, the BD permits all transfers between the computer and peripheral devices that the computer normally provides. In particular, it is usually possible to perform programmed and direct memory transfer as well as interrupt-initiated transfer. Some commercial BD's don't have provision for direct memory transfer.

The whole system is controlled by the computer program. Computer input/output instructions initiate the BD action, which, in turn, controls the branch and the Dataway operations as described.

5. CAMAC PROGRAMMING LANGUAGE

A CAMAC programming language is now in the advanced stage of development. It contains statements that reflect the characteristics of the DW and BH specifications, which themselves define the CAMAC hardware and functional details. The CAMAC language statement is a macro-instruction, which usually specifies three components: CAMAC function, CAMAC address, and computer store identifier. The control functions generally do not need a computer address. Statements may be labeled for referencing.

---

* Described and specified in EUR 4600e document.

Descriptive names may be defined and used to refer to functional entities in a CAMAC system. For example, a register implementing a counting function at a particular subaddress of a module may be given the name COUNTER. At a higher level a collection of crates may be given the name EXPERIMENT 1 and a branch may be given the name LAB 2, etc.

The general form of the statement and some examples are as follows:

label identifier	CAMAC function	CAMAC address	Computer address
GETDATA	READ	COUNTERARRAY	BUFFER
	CLEAR	COUNTERARRAY	
	WRITE	CONTROL	STATUSWORD

These examples specify the following operations:

1. Read data from a defined group of counters and deposit the contents in a buffer in the computer memory.
2. Clear the counters.
3. Load a control module with a status word specifying the details of the next run.

The examples shown represent the data transfer and control statements. There are other statements associated with the interrupt handling, status testing, etc. Basically all addressed and nonaddressed commands or functions are covered by the set of CAMAC macro-instructions. Other CAMAC statements are declarations or allocation statements. An example of such a statement is

REGISTER = (C, N, A),

where C, N, and A specify full hardware address composed of numbers for crate, station, and subaddress respectively.

CAMAC language does not include any of the facilities normally associated with data processing, such as movement of data within the memory, performance of arithmetical and logical operations, and specifications of standard peripheral activities. It must therefore be augmented by a conventional programming language such as an assembly or Fortran language.

The straightforward translator for such CAMAC language would consist of two stages. The first stage would take CAMAC language statements and produce equivalent host language statements and necessary declarations. That stage would ignore the host language statements. The second stage, the host language standard translator, would then provide the run time code.

6. AN EXAMPLE OF THE MEASURING SYSTEM IMPLEMENTED IN CAMAC

The pulse height analyzer (PHA) is one of the most accepted tools by those engaged in nuclear spectroscopy, whether a physicist researching the nature of matter or a physician involved with radiation imaging and treatment of malignant tissues.

It seems appropriate, then, to describe the implementation of a PHA system utilizing CAMAC standards. Such systems are now in operation in several laboratories in the United States and Europe.

In Fig. 8 five major blocks are seen. The spectrometer composed of a nuclear radiation detector, amplifiers and analog to digital converter is usually implemented in NIM modules.

The computer can be any of many small and medium-size computers available today. The branch driver is the only part of the system that is computer-dependent. Exactly the same system can be operated with a different computer

FIG. 8. MULTICHANNEL PHA SYSTEM IMPLEMENTED IN CAMAC

provided that the corresponding branch driver is available and that the new machine is programmed properly.

The analog-to-digital converter is interfaced to the CAMAC dataway by a transfer gate module. The basic display is operated with an X,Y,Z display driver module. No specific oscilloscope is indicated since nearly all oscilloscopes, whether of the refresh or storage type, are suitable.

Additional CAMAC modules can be used to generate character in the CRT and to operate a light pen in an interactive mode.

Console panels with switches, knobs, and dials have been found to be very convenient by many users of computerized data acquisition systems. This eliminates the need of interacting through the teletype using mnemonic words. A CAMAC system can easily interface such a console using some of the readily available modules.

Peripheral devices such as plotters and teletypewriters can also be made a part of the CAMAC Dataway but using the appropriate functional modules. Interface hardware for plotters of various types and makes as well as the ASR 33 Teletype are commercially available.

Other interfaces for magnetic tape recorders and disk drivers are also becoming available.

Although not a part of the basic PHA system, functions to control devices such as rectilinear scanners and scintillation cameras are available and can easily be added to the CAMAC Dataway.

This description of this pulse height analyzer system emphasizes the inherent benefits associated with the CAMAC standard for modular instrumentation.

7. CAMAC WORKING GROUPS AND PUBLICATIONS

After the original CAMAC standard documents EUR 4100e and 4600e were published, activities continued with CAMAC working groups, the ESONE Committee, and the NIM Committee. The following groups have been formed in both Europe and the United States: Dataway Working Group, Mechanics and Power Supply Working Group, Software Working group, and Analog Signals Working Group. Presently the major efforts are devoted to improving and revising the existing CAMAC standards and to preparing new documents, such as Analog Signals Standards and Software Standards.

The ESONE Committee publishes CAMAC Bulletin, which can be ordered from: Commission des Communautés Europeennes, D. G. XIII-CID, 29 rue Aldringen, Luxembourg. The CAMAC standard documents EUR 4100e and 4600e can be obtained in the United States from: Louis Costrell, Chairman, AEC Committee on Nuclear Instrument Modules, Radiation Physics Building, National Bureau of Standards, Washington, D. C. 20234.

ACKNOWLEDGMENTS

The authors wish to thank Charles Williams for helpful discussion and suggestions during the writing of this paper.

REFERENCES

1. CAMAC: A Modular Instrumentation System For Data Handling. Description and Specifications. EUR 4100e.
2. CAMAC: Organization of Multicrate Systems. Specification of the Branch Highway and CAMAC Crate Controller Type A. EUR 4600e.
3. T. D. Douglass, "Standard Instrument Modules," Research/Development, Vol. 22, p20 (November 1971).

4. L. Costrell, "CAMAC Instrumentation System - Introduction and General Description," IEEE Trans. Nucl. Sci. NS-18(2), 3 (1971).
5. F. A. Kirsten, "Operational Characteristics of the CAMAC Dataway", IEEE Trans. Nucl. Sci. NS-18(2), 9 (1971).
6. F. A. Kirsten, "A Short Description of the CAMAC Branch Highway," IEEE Trans. Nucl. Sci. NS-18(2), 19 (1971).
7. S. Dhawan, "CAMAC Crate Controller Type A," IEEE Trans. Nucl. Sci. NS-18(2), 33(1971).
8. M. Sarquiz and P. Valois, "An Approach to a CAMAC Language," CAMAC Bulletin No. 2, p20 (November 1971).
9. P. Wilde, "An Introduction to CAMAC: A CAMAC Language," CAMAC Bulletin No. 3, p5 (March 1972).
10. D. Sanghera, "Multichannel Analyzer System in CAMAC," CAMAC Bulletin No. 3, p9 (March 1972).

A SCANNING SYSTEM WITH TWO DATA CHANNELS,
DATA REDUCTION AND DISPLAY

P. R. Bell, W. J. McClain,* D. A. Ross,
J. K. East, and E. C. Moore

The Molecular Anatomy (MAN) Program
Oak Ridge National Laboratory
Oak Ridge, Tennessee 37830

April 1972

This work was sponsored jointly by the National Institutes of Health,
the U.S. Atomic Energy Commission, and the American Cancer Society
under contract with the Union Carbide Corporation.

---

*Computing Technology Center, Union Carbide Corporation,
Oak Ridge, Tennessee 37830.

A SCANNING SYSTEM WITH TWO DATA CHANNELS,
DATA REDUCTION AND DISPLAY

P. R. Bell, W. J. McClain, D. A. Ross,
J. K. East, and E. C. Moore

SUMMARY

A computer-based scanning system, intended for on-line operation with a dual-head rectilinear scanner is described. A simple scanner-computer interface requires only five signals to fully describe the scan for the minicomputer. Display during acquisition and immediate processing and display of the data from the completed scan permit fast evaluation both for quality of scan and for possible abnormalities. Four display formats as well as smoothing by weighted averaging, antiscatter correction by synthetic subtraction, and bounding are included in the software. Additional useful and informative routines included are a multi-contrast enhancement display which provides an immediate quantitative assessment of the data and a validator routine which specifies small but statistically sound deviations that might be considered abnormal.

A SCANNING SYSTEM WITH TWO DATA CHANNELS, DATA REDUCTION AND DISPLAY

Introduction

The computer-based scanning system is a complete on-line system for the acquisition, processing and display of rectilinear scan images.^{1,2} The system is intended to stand alone, requiring no support from a computing center. The scan image is observed in raw form as it is formed, and within 1 to 2 minutes after the end of the scan it can be processed, examined, and photographed. The usual data-reduction processes, such as smoothing and data bounding, can be completed in at most 2 minutes, while the processes of image subtraction or addition, background erase, and contrast enhancement are completed in 0.1 to 1.5 seconds. In addition, two processes not previously available are provided: (1) a program for reducing the effects of collimator septum penetration and gamma-ray scattering in the patient³ and (2) a validator program. The validator program eliminates (or strongly reduces) high or low regions of small extent (that might be small tumors) unless these small regions are found to be statistically sound by exceeding the area average surrounding them by a selected number of standard deviations. Regions thus confirmed are unmistakably flagged for investigation.

Features of the system are: (1) low cost, (2) good reliability, (3) excellent repairability, (4) simplicity of operation, (5) compatibility with clinical use with large patient throughput, and (6) no program limitation on scan width or length. The only limitation to picture size is the total number of picture elements: 16,000 in each of two images or 32,000 in one image.

Computer Requirements

A computer for this system requires 8K words of core and a 32K word magnetic disk.* The system programs all reside in the lower 4K of core at the same time; no programs are stored on the disk, which is reserved for image and comment data. The external mass storage required to keep permanent patient records is provided by any of the available inexpensive cartridge or cassette magnetic tapes.**

The reliability of the system and the small tapes is very good. We have produced over 100 separate patient images since we began using the system and have not yet lost one of them.

Signal Requirements

The system may be applied to any rectilinear scanner, single or dual head, and requires a minimum of information from the scanner. The scan line is divided

---

*We are using a Digital Equipment Corp. PDP-8/I computer with a DF-32 disk.

**At present we are using the Tennecomp TP-1351 magnetic tape with the TP-1346 autoloader.

into picture elements either (a) by a rotating slotted disk (with a lamp and photodiode) attached to the scanner; or (b) by an attached, cord-operated device that generates similar pulses. These picture-element signals are simple pulses marking the motion of the scan head through any chosen fixed distance (say $\frac{1}{4}$ or $\frac{1}{2}$ cm). The identification of scan lines is secured in either of two ways: (1) by an end-of-line signal, or (2) by a reference-line signal. The end-of-line signal occurs when the scan head reaches a chosen limit and remains until the scan head has reversed and reaches this limit traveling in the opposite direction. The reference-line signal indicates when the scan head passes a specific position near the center of the scan motion. This reference signal must occur between two picture element pulses. Either system provides sufficient information to keep the scan data in order. The second system permits either or both of the scan-width limits to be changed during a scan.

Other signals required from the scanner are, of course, the stream of pulses from the spectrometer attached to the detecting head, or two such streams from a dual-head scanner. These streams of data pulses may, of course, result from two different energy windows which permit detecting counts from two separate radionuclides or from two energies of one nuclide. Finally, a landmark signal, though not really mandatory, has been found extremely useful. A photodiode is inserted in the central hole of the detector collimator, and it views small low-power light bulbs attached to anatomical bench marks on the patient. These landmark signals are detected during the scans and are recorded along with, but separate from, the gamma-ray pulses. They may be turned off or on in the displayed image as desired.

These 4 signals (or 5 for a dual-head system) are transmitted to the computer for processing; no precounting or other processing is performed.

System Operating Modes

The computer programs operate with the program interrupt enabled: that is, the computer will respond to an external request from the scanner interface or from the teletype. The programs have been grouped functionally into two main modes: (1) the data-acquisition mode, and (2) the data-handling mode. Operation in the former mode may be interrupted by signals from either the scanner or the keyboard; in the data-handling mode the program can be effectively interrupted only from the keyboard.

An elementary keyboard monitor responds primarily to single-letter or single-letter-and-number commands. Processing routines and a scan-record-entry routine are called from the data-handling mode by the two-key command CTRL-C followed by the program name. Specific commands are acceptable in each of the two modes, but each mode responds only to commands contained in itself. Should an inappropriate (or any non-command) character be given, a question mark is typed and the command is ignored.

Transfer of control between the two command modes is actuated by the two-key commands: CTRL-P (which activates the processing mode) and CTRL-A (activating the access mode). This transfer can be made at any time. Commands available within each of the two modes are listed in Tables 1 and 2.

Data Access

When the scanner is started, the computer program is started by typing G (for GO) on the keyboard. The computer begins to count the picture elements after the first end-of-line signal appears. When the next end-of-line occurs, the computer knows the chosen line length; it then begins to retain the data and to display them in raw form. If the reference-line system is being used, the occurrence of two reference-line signals permits the first two half lines

TABLE 1

COMMANDS FOR DATA ACCESS MODE

<u>COMMAND</u>	<u>ACTION</u>
G	GO, starts ACCESS-1
Dn	DISPLAY level, n = ϕ -3
Q	QUIT, proceed to REARRANGE
In	ID, n = decimal identification number
U ϕ	Selects lower image
U1	Selects upper image
C	COMMENT; enter, add to or change comment
CTRL-P	Transfer to Data Handling Mode

TABLE 2

COMMANDS FOR DATA HANDLING MODE

<u>COMMAND</u>	<u>ACTION</u>
--- DISPLAY COMMANDS ---	
Dn	DISPLAY level, $n = \phi-7$
Sn	SUBTRACT, n = decimal number to subtract
L	LANDMARKS, show if off, turn off if on
En	EXTEND display size
Pn	PHOTO, provides n sweeps of display
Cn,Cm	AREA CONTOUR, displays counts from n to $n+m$ inclusively
Cn	LINE CONTOUR, displays a contour line at n counts
M	MCCE, multicycle contrast enhancerent
Vp	VARUL, change m to $m-n$ in AREA CONTOUR (or lower level by 1 count in LINE CONTOUR) after p sweeps
X	FLIPX, reverses X coordinate
Y	FLIPY, reverses Y coordinate
F	FORWARD, display next 4K block of data
B	BACK, display first 4K block of data
T	TOTAL, display all 4K blocks of data
Zn	Z-PROFILE, display profile in line n
In	ISOMETRIC display, all blocks. Rotation controlled by n .
--- MAGNETIC TAPE INPUT-OUTPUT ---	
Rn	READ four 4K blocks from magnetic tape; n = track if on AUTO
Wn	WRITE four 4K blocks on magnetic tape; n = track if on AUTO
#n	REFRESH complete disk from track n of magnetic tape
\$n	DUMP complete disk onto track n of magnetic tape
--- DISK OPERATION ---	
An	COPY, $n = \phi$, upper to lower disk; $n = 1$, lower to upper disk
Nn	SELECT disk area; ϕ = lower, 1 = upper
--- DATA MODIFICATION AND QUANTITATION ---	
CTRL-Cname	Call to various processes by name
ϕ	LITEUP, show quantitation marks
ln	Set line of first mark to n
2n	Set element of first mark to n
3n	Set height of rectangle to n

contd.

TABLE 2 - contd.

<u>COMMAND</u>	<u>ACTION</u>
$4n$	Set width of rectangle to n
Kn	Operation on data within boundary marks $n = \emptyset$: sum all counts $n = 1$: list all counts $n = 2$: type histogram of area
--- PROGRAM ACTION ---	
Q	QUIT, abort operation in progress and wait
CTRL-A	Transfer program control to data access mode
--- COMMENT ENTRY ---	
CTRL-CNOTE	Enter, add to or change comments
CTRL-CPUT	PUT comment into image file in current N state
--- COMMENT MODE ---	
CTRL-K	KILL comment
RUBOUT	RUBOUT one character
CTRL-FORM	RETURN to command mode

to be displayed. At each succeeding passage of the reference line, information is available for the correct display of the preceding excursion of the scan from reference line to edge and back to reference line. As the scan grows element by element, the display matches it point by point.

This image during access is invaluable for ensuring correct patient placement and correct operation of the scanner. The intensity of the image may be adjusted during the scan by keyboard commands without altering the recorded data in any way. When two-detector (or two-energy) scans are being made, the operator may shift the display from one image to the other, or vice versa, whenever he desires.

When the scan is completed, a Q (for QUIT) command is typed and the scan data files are immediately reordered from the boustrophedonic format into array format. This done, the computer requests a scan identification number, which is supplied by keyboard. The scan data are immediately displayed and are ready for processing or storage.

Data Processing

The data-processing facilities of the system are all intended for quick and immediate action. Generally, the data in raw form are dumped onto a small magnetic tape cartridge for later recovery if required; the data images in the computer can then be processed at once. The matrix averaging program or the antiscattering combined with matrix averaging require roughly 1 to 2 minutes, depending on the number of picture elements and the total count content. A large scan of about 16,000 elements requires approximately 2 minutes 10 seconds for processing. The antiscattering without smoothing, or the bounding programs alone, require about 1 to 2 minutes, but since these processes produce "raw" (that is, unsmoothed) data, they must be followed by matrix averaging which requires an additional 1 to 2 minutes. Complete validation similarly requires more total time since it can be applied only to smoothed data; the validator program itself requires about the same time as the averager.

Dual-input scans can be added together or subtracted, picture-element by picture-element, and this allows, for example, the formation of a composite posterior-and-anterior view of the patient. Or, in a patient who has received both technetium sulfur colloid and selenomethionine, it permits the subtraction of the liver image from the liver-plus-pancreas scan, thus unmasking the view of the pancreas.⁴ The addition-subtraction process is quick, requiring only 5 to 10 seconds, but some time may be used in selecting the proper multiplying factors using preliminary trial images (formed in, at most, one second). The full data file can then be committed to the process.

Data Display

Scanning systems have previously been plagued by the quality of their output, the hard copy of the scan image. One-shot records such as from dot-tappers and photo-recorders do not usually provide the quality required for easy interpretation of the scan, and, of course, the data are not recoverable for processing. Intensity mapping using symbols on a line printer produce images of poor quality with rather bad resolution.

The optimum display system is one that (1) avoids destruction of the basic data, (2) provides a copy without excessive delay, (3) is inexpensive, and (4) provides an image that is smooth and easily interpreted. Photographic copies of CRT displays having interpolation approach these optimum requirements. The grey scale of Polaroid film is adequate for scan interpretation if more than one image is available and avoids the risk of finding a bad picture of the record after considerable delay for developing. Variable scaling, background erase,

and various display formats may be used to provide multiple views of the same data with the views varying in detail and thereby emphasizing different features of the image. Interpretability may be enhanced by use of different views of the same data averaged by coarse and sharp averages. In this manner, bumps due only to statistics can be recognized and discounted in the interpretation. This requires quick recall of the basic data for multiple processing and viewing. Variable-resolution averaging⁵ provides a difficult recognition problem since statistical fluctuations (lumps) have different sizes in different parts of the same image.

Contrasted with data processing, display manipulation is very fast. Some of the manipulations available are: changes in background erase level, contrast enhancement factors, image size, orientation or image type (intensity map, contour, profile, multicycle contrast enhancement, or isometric). Changes are completed as fast as the commands are typed. The swiftness of change allows rapid choice of proper display constants for enhancement of any part of the image for optimum photography. Once the desired display parameters have been selected, the photo command (P) ensures preselected exposure for photography.

Examples of the display types available are shown in Fig. 1. The standard intensity map [Fig. 1(a)] and the multicycle contrast enhancement display [Fig. 1(b)] are interpolated displays; that is, the number of points displayed is four times as large as the number of data points in core. Points between data points are averages of adjacent points. This yields an exceptionally smooth and readable picture. Figure 2 shows a comparison of interpolated and non-interpolated images.

Two types of contour plots are available, except that only one type is present in the program at one time. The line contour [Fig. 1(c)] marks a single point on each scan line at every point at which the successive data values pass up through or down to the commanded level. The area or zone contour [Fig. 1(d)] intensifies all data points that lie between the commanded level and that level plus a commanded difference. Each type of contour plot may be caused to regress from high to low level automatically.

The isometric plot [Fig. 1(e)] is a two-dimensional representation of a three-dimensional figure where activity or count value is plotted as the third dimension. These images may be rotated to right or left by keyboard command.

The Z-profile [Fig. 1(f)] is a plot of the count value vs. position for the data points along a commanded scan line. This provides a rapid means of examining the quality of the scan data.

Provision is made for the addition of other display modes as their need may arise.

An important feature of the displays is the special marks appearing in them. The landmarks denote the automatically recorded location of small lamps that were attached to strategic points on the patient during the scan [see Fig. 1(a)]. Similar marks are provided by the validator program denoting points that represent statistically valid deviations from the surrounding region. Both types of marks are commanded on or off in the image by typing L. Another type of mark is inserted and moved in the image by the quantitation commands. These points mark the corners of a rectangle whose size and location are keyboard commanded. The program can then provide a sum of the enclosed data points, a type-out in array format of the data points, or a histogram of the data content. Since the markers have digital locations, they may be placed similarly in the other image of a pair, or at any desired area in the same image for comparison purposes.

Fig. 1. Six types of display available with this program. The standard intensity map with optional landmarks is shown in (a). The multicycle contrast enhancement image is shown in (b). Line and area contours are shown in (c) and (d), respectively. The isometric view appears in (e), and a Z-profile display of a single line across the scan is shown in (f).

We have found that the multicycle contrast enhancement display provides semiquantitative images that are useful when used with a standard intensity display. Figure 3 shows a left lateral brain scan with a standard display and a multicycle display. A background erase corresponding to 17.5 counts/element was used in both. Inside the head the intensity rises through only one cycle of display over most of the brain area. This corresponds to an increase of 8 counts/picture element. As it approaches the high activity region near the vertex, the image switches to black as the first cycle is exceeded. The image then gradually brightens again through the next display cycle and again switches to black at the hot spot at the vertex. This corresponds to another increase of 8 counts/picture element. The vertex hot-spot region thus corresponds to $2 \times 8 + 17.5$ or about 33.5 counts/element. The intensity value at any point is thus easily appreciated without having to detect an "image" represented by a field of numbers. The high contrast enhancement provided makes for easy visualization of the normal pattern.

An example of the performance of the antiscatter program is shown in Fig. 4. Figure 4(a) is a raw data image of a Ga-67 scan. Integration of the point-source response of our system shows that about 50% of the response lies in the very low but very wide skirts. This is almost all scattering in the patient since septum penetration is very small in the gold collimator used. Figure 4(b) shows the "raw" data resulting after the contamination produced by the skirt of the point-source function was calculated and subtracted. Figures 4(c) and 4(d) show the smoothed images of these two raw-data images. Note the considerable reduction of the scattering haze.

Figure 5 illustrates the performance of the validation program, which calls attention to small but statistically sound deviations that might otherwise be missed by the viewer. A clinician must, of course, judge the biological significance of such deviations. Figure 5(a) shows the raw scan of a patient with Hodgkin's disease, and clearly there are numerous hot spots. Figure 5(b) is the averaged version of the same scan. In Fig. 5(c) the validator is indicating the areas where the activity deviates from the surrounding average by 1.4 standard deviations or more. In Fig. 5(d) the threshold deviation is 2 S.D.; the marked areas are therefore fewer in number but of greater significance. Figures 5(e) and 5(f) show profiles of one scan line through the hottest spot: the one high in the thorax, just to the left of the midline. In Fig. 5(e) the points at the top indicate the positions of the picture elements where the count exceeded the threshold deviation (2 S.D.); directly below these dots the lower curve shows the deviating values. Elsewhere in the lower curve the original picture-element counts have been replaced by the flat averages determined by the validator. Figure 5(f) shows a profile for the same scan line, after the averaging performed in Fig. 5(b) but without further processing by the validator.

Fig. 2. The effect of interpolation during display. The image in (a) contains 4071 displayed points without interpolation while that in (b), using the same data, contains 16,146 displayed points. The subject imaged is a 280 g rat with a 0.28 mCi dose of Ga-67, scanned after 24 hours. Despite the fact that (b) contains no more information than (a), it is much easier to interpret.

Fig. 3. Left lateral head scan. The image in (a) is the standard intensity map while that in (b) is the multicycle contrast enhancement image of the same data. From view (b) it is easy to note that the parotid gland activity is somewhat more than 48 counts/element above the activity at the lowest point in the brain (6+ cycles). Scanning time for this scan was 15 minutes.

Fig. 4. Examples of the results of the antiscattering program. The raw data is shown in (a). Figure (b) shows the data after antiscattering correction, but still in unaveraged form. View (c) shows the same data averaged by the 7×7 averager. Figure (d) is the data of (b) averaged by the matrix averager. Note in (d) the considerable reduction in haze seen in (c). Lower left (e) is a superposition of a single line through both (c) and (d) at the location of the xiphoid process landmark. The upper or wider set of points is that with no antiscattering correction. The lower right (f) shows a similar superposition from (c) and (d) at the level of the left shoulder landmark.

Fig. 5. Results of the validator program on a difficult clinical example. View (a) shows the raw data, and (b) shows the same data averaged by the 7×7 averager. The view in (c) is the data of (b) after validation with a 1.4 standard-deviation threshold. Too many points exceed the limits and are marked. The data were double-averaged and then validated at 2 standard deviations [view (d)]. In view (e) profiles of one scan line after the validation of (d) are shown. This line passes through the hot area located high in the thorax just left of the midline. The points at the top indicate the positions of picture elements where the counts exceeded the threshold. Directly below these are the values of these counts retained by the validator. Elsewhere in (e) the original counts have been replaced by the flat averages determined by the validator. View (f) shows a profile of the same scan line after averaging [as in (b)] but without validation.

REFERENCES

1. P. R. Bell et al., "A Practical Computer-Based Rectilinear Scanning System," ORNL-TM-3652, Dec. 1971.
2. W. J. McClain, P. R. Bell, and D. A. Ross, "A Keyboard-Controlled Computer-Based Rectilinear Scanning System for Nuclear Medicine," Proceedings of the 10th Annual IEEE Region 3 Convention, April 10-12, 1972.
3. P. R. Bell and W. J. McClain, "Processing and Display of Radioisotope Scans With a Small Computer Using Synthetic Subtraction," Proceedings of Symposium on Sharing of Computer Programs and Technology in Nuclear Medicine, Oak Ridge, Tennessee, April 2-3, 1971.
4. J. J. Erickson and A. B. Brill, "Physical Considerations for Dual Radio-nuclide Imaging Techniques," in Quantitative Organ Visualization in Nuclear Medicine, Edited by P. J. Kenny and E. M. Smith, pp. 777-820 (University of Miami Press, 1971).
5. Stephen M. Pizer, "Digital Spatial Filtering and Its Variations," in Quantitative Organ Visualization in Nuclear Medicine, Edited by P. J. Kenny and E. M. Smith, pp. 581-603 (University of Miami Press, 1971).

THE GAMMA-11 SYSTEM, A NEW TOOL IN THE NUCLEAR MEDICINE DEPARTMENT

John Silber

Digital Equipment Corp.
Maynard, Massachusetts

ABSTRACT

A new data acquisition, processing, and display system for Gamma cameras is described.

The system is oriented to clinical operations but is powerful and modular enough to permit research in advanced nuclear medicine techniques.

Fast acquisition, sophisticated processing and labeled displays are features of the system. Equipment used is a PDP-11, with disk cartridge storage, display tube, and DEC writer. Operations are simple to learn and use since questions appear in English on the screen and are answered in the same way by the operator.

April, 1972

THE GAMMA-11 SYSTEM, A NEW TOOL IN THE NUCLEAR MEDICINE DEPARTMENT

by John Silber
Digital Equipment Corp.
Maynard, Massachusetts

What is GAMMA-11 ?

In the last few years many universities in this country and abroad have invested time, money and effort in developing a variety of computer systems to collect and analyze gamma camera data. Some used DEC computers, the PDP-8, 12 and 15, others worked with IBM's, Computer Automation and other machines.

The results were encouraging and led us to believe that a computer pack synthesizing the best features of each would be useful to the Nuclear Medicine specialist.

In analyzing this pioneer work, we reached some conclusions on which we based our design:

The 12-bit word length is too large for most dynamic studies and too small for statics. The half word (6 bits) is too small for slow dynamic studies. Thus, we chose an advanced 16-bit machine, the PDP-11 and used the half word as the element size for dynamic studies. This provides maximum core and disk storage efficiency.

Magtape and DECTape had transfer rates we considered low for the maximum dynamic study data rate. Thus, we chose disk as the primary storage medium, with tape backup.

The refresh display with or without spot modulation did not provide a satisfactory picture because too slow (flicker-full) or too "spotty".

We chose a storage tube that enables us to get a good scale of gray and characters simultaneously. The characters describe the patient and the study parameters.

The computer is not considered as an appendix to the camera, another instrument. GAMMA-11 provides full patient data storage and can accommodate the physician's clinical comments along with images, time/activity curves and all related information. The system makes the transition from image enhancement, data processing

and display to full utility as a clinical unit in the Nuclear Medicine department.

At off-hours the hardware can be put to other uses by running MUMPS, a clinical data management language. Typical applications are patient files, admissions, radio-pharmaceutical inventories, etc.

Generally, the system is oriented to the clinician but its capability is more than enough for the most demanding researchers. Full, flexible disk programming is present in machine language, FORTRAN, BASIC and FOCAL. The complete range of PDP-11 peripherals is available.

The Camera, the System and the Interface

The three Anger type cameras on the U.S. market today operate on the same principle but generate pulses of different shapes and duration. Furthermore, attachments can be purchased for dual isotope studies or tomography.

Some cameras are supplied with integral videotape storage devices including ADC's, others offer such equipment as add-ons. Usually regions of interest may be summed with this equipment but only very few of them and only roughly. Also, the videotape loses 5 - 7 % of the data in the recording process. It takes as long to obtain the two regions of interest data as it took to record it.

The output is photographed on a Polaroid camera with three diaphragm openings to insure at least one image with good contrast. The name of the patient, file number, view and other pertinent data are scratched on the photo by hand.

In contrast, once the computer has acquired the data many operations may be quickly performed on the array or the display. The image is labeled with the appropriate patient data and confusion disappears.

Data analysis is performed quickly and no hand calculations are necessary: results are printed out or displayed as desired.

Patients may be accessed in a fraction of the time needed to spin a videotape to the right place. The operator chooses the patient needed from an index on the screen and this data is displayed in about one-half second.

The interface connects to the camera on the x, y and z leads, bipolar or unipolar as desired. On a Pho Gamma, we pick up the +x, -x, +y, -y, z leads after the rotation and orientation switches, on the A xcope. If the N-C Data Store is present we

can pick up after the summing amps on the B scope, enabling replay from the storage unit. On a Dynacamera, pickup is before the ADC because of its long conversion time. On a dual isotope N-C unit, the $+x_1$, $-x_2$, etc., are summed before the ADC in the interface. The z pulses are segregated and trigger appropriately. The ADC's used in the GAMMA-11 interface are 50 MHz specially made for DEC, converting to 7 bits in 5.2 μ s including settling times.

The interface has three stages of digital buffering slaved to each other: the address progresses down the chain only as each level is freed through processing. The maximum time for processing three pulses is about 24 μ s in the worst case. Our calculations show that the loss rate is about 1.6% at a maximum realistic count rate of 30,000 cps or about .8% at 20,000 cps.

An increase in ADC clock rate would not improve these loss rates: a 100 MHz ADC must be run over 8 bits because of inherent imprecisions at these extremely high rates. Thus, 100 MHz at 8 bits is equivalent to 50 MHz at 7 bits. Two hundred MHz ADC's are quite imprecise and tempermental. If higher speeds are desired we prefer to increase the digital buffer by one or two stages. Analog buffers before the ADC are also possible.

The interface collects in frame or list mode, program selectable. Also program selectable are matrices, inscribed and circumscribed to the crystal face. List mode data can be run through the interface again to be converted to frames at high speed.

The system consists of a PDP-11/20 computer with 16K of 900 ns. core. The disk used is of the moving head cartridge type and stores 2.4 M bytes. To back up the disk a fast paper tape reader and punch is present.

The display scope used is of the storage type, Tektronix 611. Its point plotting speed is 100 μ s/point and has a resolution of 2048 x 2048. The interaction is conducted between the DECwriter keyboard and the scope. The printer part of the DECwriter is used only on request, for hard copy.

Acquisition

A programmable control register and offset register in the interface can be set by the programs to determine where and in what mode the data is to be stored in core.

Matrix sizes of 32 by 32, 64 by 64 are possible with the element size being a byte or a word. A 128 by 128 byte matrix is also available for high resolution static studies.

The acquisition frame rate and resolution are linked and

and may be matched for best speed and storage efficiency: for example, if counting at 20,000 cps, one may collect 64 by 64 by 16-bit frames at 7/second.

But, the depth of the bin or element, is wasted and the camera resolution often does not warrant a 64 by 64 matrix.

The rates are:

32 x 32 x 8	60 frames/sec.
32 x 32 x 16	30 frames/sec.
64 x 64 x 8	15 frames/sec.
64 x 64 x 16	7 frames/sec.

There is no dead time between frames as each is buffered in the upper 12K of core: one area acquires, the other dumps on disk, then they reverse. In list mode the maximum count rate is above 30,000 cps and the reframing can be done at 100 frames/sec.

The acquisition programs are set up in question and answer mode by answering multiple choice questions or entering patient data, item by item. All data may be edited and reviewed before the study proceeds. When this is completed, a key is struck or the remote switch thrown for acquisition to start.

When acquisition is over, the index is updated with the patient entry, ready for selection and display.

The disk is large, but if long-term storage of data is needed, it's best to have magtape for backup. An indexing scheme allows the user to page through the tape and select a study to be viewed. The tape is also needed if a MUMPS capability is desired.

Processing

The images are stored on disk with index entries and their patient data appended. A special 500 character block is set aside for physician comments at the end of the study.

The review process starts when the operator presses the appropriate key and views the index on the screen. The patient entries are numbered and appear in reverse chronological order. Each entry gives the patient name, file number, type of study, site, length, data, etc. When a patient is selected the first image in a dynamic study or the static image appears. Processing depends on the type of study.

Static Processing

The image may be flood corrected to compensate for crystal

inhomogeneities and correct for the camera transfer function. Several kinds of smoothing routines may be applied: matrix smoothing (1), 5 point, 9 point, variable spatial, etc. As new smoothing or enhancement methods become accepted, they will be inserted in the system. Thresholds are chosen interactively via the joystick on the screen. The thresholding eliminates all cells below and above the set level, spreading the scale of gray linearly or nonlinearly, over the remaining count band, thus providing a form of edge enhancement.

Vertical and horizontal slices may be displayed one at a time or in groups via joystick interaction. Frames may be added, subtracted, multiplied and divided at will.

When the image is ready for photography it is redisplayed showing patient data and all operations performed to that point. This is necessary to insure that only images processed in the same way are compared visually. The Polaroid camera supplied with the system reduces the scope image by a linear factor of 2 (or surface factor of 4).

Dynamic Processing

The dynamic study may at first be considered as a series of static frames and treated as such. Dynamic processing consists of a dead time correction and the selection of regions of interest. This latter process is effected via the joystick controlling a bright spot on the screen. The spot is detented from center cell to center cell and is fixed by pressing a switch on the joystick. The area can be chosen in a continuous mode or cell by cell, on the boundary.

Up to 12 areas are available and they may overlap. There are also predefined ways of obtaining regions of interest: circles, rectangles, etc.

The time activity curves are then displayed, properly labeled, on the screen. The curves may also be numerically processed using a FOCAL-like language: smoothing, curve fitting, etc.

All acquisition and processing may be predefined². For example, if a routine heart study consists of 10 32 x 32 x 8 at 10/second and 10 64 x 64 x 8 at 1/second, etc., the user may define this as a BRAIN and call it on the screen as such. Same can be done in processing, thus saving the time needed to preset the study every time. This also provides standard photos that can be compared from patient to patient.

Future

The system described is only a beginning since much work

(1) P. R. Bell, Oak Ridge National Laboratory

must be done in solving actual clinical problems. Various people are working on cardiac shunts, ventilation-perfusion studies, heart output measurements, clearances, etc. These will become available as soon as accepted by the Medical profession and constitute a true step forward in Nuclear Medicine data processing.

Furthermore, there are several institutions involved in developing a radiotherapy package on the PDP-11 to be run at off times.

Conclusion

GAMMA-11 provides the clinician with an easy to use system putting all gamma camera data at his fingertips. Its facility in setup and analysis will soon make it a valuable quantitative adjunct to the basic camera.

BASIC CONSIDERATIONS IN COMPUTER USE FOR
DYNAMIC QUANTITATIVE RADIONUCLIDE STUDIES

Robert H. Jones, M.D., Bryce B. Bates, B.S., Jack K. Goodrich, M.D.
and C. Craig Harris, M.S.

ABSTRACT

Over the past several years, use of stationary detectors interfaced with a computer for data storage and manipulation have generated widespread interest in dynamic radionuclide studies. A major limitation for these studies is the long event resolution time inherent in all current instrumentation that limits the total count rate to about 25,000 counts per second. In spite of this limitation, a number of studies have suggested that the innocuous nature of tracer technics for dynamic studies hold great potential benefit for patient management. A general approach is described which is applicable to all dynamic isotope studies. An initial step assigns digital counts to discrete time and space intervals. Data is corrected for distortion from instrument dead-time loss and for variation in detector efficiency over the field of view. Detector regions which correspond to anatomic structure must be accurately delineated, and characterizations of the resulting data which reflect regional organ function are determined.

From Department of Surgery
and the Division of Nuclear Medicine
Duke University Medical Center
Durham, N. C. 27710

Introduction

Computers have been used in nuclear medicine primarily for contrast enhancement and modification of pictorial display of static images of radioactivity.^{1,2} Following initial studies in 1967, widespread interest has gradually developed in computer technics for dynamic radionuclide studies.³ Rapid, sequential recording of changes in the distribution of radioactivity requires use of a stationary camera-type detector. Although important physiologic observations have resulted from many initial studies, no dynamic radionuclide study of regional organ function has developed into a standard clinical procedure. The major detriment to the accuracy and usefulness of this approach results from limitations of available instrumentation. Electronic dead-time of current instruments severely limit the maximum counting rate. However, current efforts to produce instruments with high counting rates may overcome this limitation. Increased instrument performance should stimulate further radiopharmaceutical development with a particular impetus for development for ultra short-lived radionuclides to achieve high resolution dynamic studies. With increasing use of dynamic radionuclide studies, standardization of technics appears important to assess the ultimate role of the information obtained in patient management. This report describes a general approach which is applicable to all dynamic quantitative radionuclide studies. Performance characteristics of available instrumentation are summarized, and technics for data accumulation, correction, manipulation and display are described.

Instrumentation

The need for a rapid static image of the distribution of radioactivity within patients provided the major economic impetus for commercial development and production of stationary detectors. Secondary adaptation of stationary imaging devices to regional dynamic studies has usually followed the least expensive course and therefore has produced instrumentation with only minimal performance characteristics. Basic principles essential for accurate detection of static radioactivity are equally important for dynamic studies. Factors such as collimator design, pulse-height resolution, accurate event positioning, and detector field uniformity influence the accuracy of dynamic data in a manner similar to static imaging. However, the low count rate possible with current detecting devices represents the greatest limitation for dynamic quantitative studies.⁴ The electronic dead-time required for individual event processing represents the primary cause of low count rates achieved. Use of the term dead-time leads to inaccurate comparisons among available instruments because of the complex multi-component nature of this parameter. A more accurate and practical term might be event resolving time which can best be indexed by the maximum number of digital, spatially oriented counts per second available from the stationary detector. Commercially available instruments become inaccurate at counting rates greater than 25,000 counts per second. The severe limitations of low counting rates are well illustrated by radionuclide angiocardioqram data. In these studies, optimal time intervals are 0.2 seconds or less, and each information element should be no larger than 1 centimeter square. Radionuclide angiocardioqram data obtained with the Digital Autofluoroscope, an instrument which consistantly produces a total of 25,000 counts per second, demonstrates maximum counts of only 200 in the area of injection

of 10 millicuries of Tc-99m pertechnetate into the right ventricle (Fig. 1). The time and space separation used permits an average of about 16 counts per information unit, and higher counts are observed in the right ventricle only because counts are low in the surrounding crystals. The limitation of count rate becomes even more apparent in the left heart phase after the bolus of radioactivity passed through the central circulation and exposed a larger number of detector units to radiation. The data distortion caused by event resolution time can be corrected, but the statistical limitations imposed by count loss can be overcome only by improvement of instrumentation to provide greater observed count rates.

Presently available commercial systems for dynamic radionuclide quantitation are summarized in Table 1. Detectors which may be used include either a single-crystal gamma camera or use of a mosaic of individual scintillation crystals. The availability of a gamma camera in most departments of nuclear medicine has made attractive the interface of this detector with a small computer for dynamic radionuclide studies. Although certain unique features characterize each system, common components include the gamma camera with a rapid analog to digital converter and a buffer storage to aid rapid introduction of count information into the central processing unit.⁵ Most data processing systems permit accumulation of about 300,000 counts per second and the major component of instrument count loss resides in the camera. The electronic dead-time of the best available gamma cameras can be measured electronically to range from 5 to 10 microseconds. The performance of the most rapid cameras when interfaced to a digital data retrieval device can best be described by a dead-time of about 15 microseconds. However, the count resolution time of the camera is a complex function and distortion of event positioning appears to limit counting rates to about 20,000 counts per second.⁶ Unsubstantiated reports have suggested maximum gamma camera count rates of about 40,000 counts per second, but no published studies have confirmed accurate event resolution at counting rates exceeding 25,000 counts per second.

Use of multiple individual scintillation probes provides an alternate approach for regional quantitation of dynamic isotope data.⁷ This apparently simple approach has found limited application because of the difficulty in obtaining a concomitant image of observed radioactivity. The crystal matrix can be designed to fulfill the needs of individual studies, and few as four individual scintillation probes with a computer retrieval of data has been observed to provide useful information in cerebral circulation studies.⁸ The Digital Autofluoroscope is a commercial instrument with a multi-crystal detector which provides a computer interface through magnetic tape recording of data. Unfortunately, the 294 scintillation probes of the Autofluoroscope are not electronically independent and the entire crystal matrix responds as a single unit in dead-time generation. Therefore, a count observed in one crystal determines that an event will not be recognized in any other crystal for a 13 microsecond period. Current autofluoroscope design limits the total matrix counts to a maximum of 25,000 counts per second, and the entire system operates with an effective resolution time of 24 microseconds. Despite presently observed limitations, an instrument with a mosaic of many electronically independent probes with direct

2 Seconds Right Ventricle

1	1	0	8		6	9	3	8	12	8	11			
4	1	3	7		6	11	17	20	24	29	33	19	14	
5	4	7	6	15	20	28	27	32	67	47	60	32	23	7
8	14	12	12	14	22	27	38	41	100	91	118	94	20	13
17	6	16	14	2	14	24	40	46	166	181	204	112	21	11
1	6	6	1	8	11	7	18	36	82	130	157	127	98	17
2	1	2	4	10	6	15	10	13	34	86	70	103	47	27
0	2	0	1	2	5	7	7	9	16	17	39	22	26	7
	2	0	1	1	5	2	8	6	6	12	9	14	19	3
	0	2	1	2	2	4	4	10	4	8	9	7	4	6
	0	1	3	0	0	0	4	5	3	7	5	4	4	2
	1	0	1	3							2	3	2	5

AVERAGE BACKGROUND 3 COUNTS PER CM.

10 Seconds Left Ventricle

9	4	4	8		15	7	5	7	13	9	6			
6	15	11	20		21	16	5	9	14	16	9	6	7	
11	23	21	30	14	28	12	12	9	13	3	13	3	6	4
22	27	18	37	34	21	34	21	4	12	16	4	9	2	1
34	43	40	57	25	42	42	22	7	13	13	16	8	2	5
15	31	24	7	47	50	43	43	48	17	30	7	3	13	6
20	35	31	20	21	33	50	40	30	23	27	16	14	3	12
15	25	17	6	23	31	30	36	28	21	29	29	13	18	11
	24	11	17	20	24	37	31	47	44	30	31	39	23	27
	8	13	8	25	19	35	33	25	22	43	58	26	37	32
	8	13	18	27	13	15	17	16	33	29	41	31	42	39
	11	9	13	15							23	34	13	23

AVERAGE BACKGROUND 9 COUNTS PER CM.

Figure 1. Radionuclide angiocardigram data at 0.2 second accumulation intervals illustrate the limitation imposed by instrument dead-time. In this study using the Digital Autofluoroscope, direct right ventricular injection of 10 mCi Tc-99m pertechnetate produces a maximum 0.2 second count of 204. This rate of over 1000 counts per second in a square centimeter represents the highest count rate which might be expected in biologic studies using this or other instruments with maximum counting rates limited at 25,000 per second. Data observed at the 0.2 second interval 10 seconds following injection when the left ventricle counts were greatest demonstrate a maximum count of only 58. This lower count rate results partially from spread of the radionuclide bolus but also reflects greater dead-time count loss caused by exposure of a larger number of detector units to radiation.

TABLE I. GAMMA CAMERAS AND CAMERA-COMPUTER SYSTEMS

Corporation	Instrument	Design	Dead Time uSec	Max* K Ct /Sec	Cost° x \$1000	Compatibility
1. Nuclear Chicago	Pho Gamma HP	12" Crystal, 19 PM tubes	5 (A) 10 (D)	20	49	7,8,9,10,11
2. Conuclear	Icon II Icon 380	13" Crystal, 19 PM tubes 15" Crystal, 37 PM tubes	5 (A)		40 60	7,8,9,10,11
3. Nuclear Data	Radicamera 50	13" Crystal, 19 PM tubes	5 (A)		42	7,8,9,10,11
4. Picker	Dynacamera 3	13" Crystal, 19PM tubes, region of interest	12 (A)	10	60	6, Probably 7,8,9,10,11
5. Baird-Atomic	Auto fluoroscope	294 1 cm crystal, tape storage Region of interest	13 (D)	25	92	Tape to large computer
6. Picker	JES	12 mHz A-D Buffer, PDP-8 Computer tape	32 (D)	10	49	Tape
7. Intertechnique	Model 66	A-D, No Buffer, 4K Computer, tape	5	100	56	Tape
8. Nuclear Data	Med II	100 mHz A-D, Buffer, 16K x 12 Bit Computer 10 megabit disk		100	45	Optional tape, Hardwire interface to lg. computer
9. Digital Equip.	Gamma II	50 mHz A-D, Buffer, 22 megabit disk	6	100	56	Optional tape, Hardwire interface possible
10. Hewlett Packard	5407A	200 mHz A-D Buffer, 12K x 16 Bit Computer	3	300	50	Tape, Optional disk, Hard- wire interface possible
11. Medical Data Systems	NMC	A-D Buffer, Nova 1200 CPU-12K x 16 Bit 48 megabit disk		400	59	Disk, Optional tape, Hard- wire interface possible

*Total count rates are limited by the slowest system component. High count rates of data systems mentioned have not been observed with a gamma camera interface.

°Cost varies with options and more detailed cost comparison would be required to assess values.

Note: Honest effort has been made to provide factual data. However, paucity of published observations describing instrument performance has forced reliance on manufacturers marketing information which may vary in reliability.

computer input for data storage and manipulation represents a promising approach for future development of instrumentation for dynamic radionuclide studies.

In summary, low counting rates possible with available instrumentation represents the major limitation for dynamic radionuclide quantitation. A major consideration for instrumentation comparison is the maximum counting rate which is achieved with accuracy. No current system holds distinct advantages and the choice of instrumentation at present is determined primarily by secondary design features which might facilitate specific dynamic radionuclide studies planned.

Approach to Dynamic Quantitative Studies

The ease of observing radionuclide transit in patients injected for static imaging procedures and the desire to use newly acquired instrumentation has prompted accumulation of a large amount of quantitative radionuclide data of questionable validity and usefulness. The amassing of a quantity of data which has not been carefully recorded and processed in well designed and controlled studies has fostered disillusionment regarding the overall usefulness of dynamic radionuclide procedures. A quantity of data cannot provide useful information without a careful approach for data acquisition and processing. Although details of dynamic radionuclide studies vary with different organ systems and radiopharmaceuticals, general principles are common to all procedures. A general approach to dynamic quantitative radionuclide studies is outlined in Table II. Digital counts must be recorded in discrete units of time and space. Data comprising a unique study should be properly edited, labeled, and stored for recall. Observed counts must be corrected for instrument dead-time loss and for variation in efficiency over the detector surface. Information points are grouped into regions which correspond to anatomic structure and the count-intensity curves obtained from these areas are used to describe physiologic functions.

The time and space resolution required by the specific dynamic function study determines the type of instrumentation required to complete the study. Some commercial instrumentation contains limited intrinsic capacities for manipulation of dynamic quantitative data which may prove sufficient for certain studies. More complex studies can be completed by adding a small central computing system with software for computer manipulation of data. In more complicated dynamic function studies such as radionuclide angiocardiograms, the memory capacity of a large computer provides a distinct advantage for complex data analysis. Regardless of the magnitude of computations required to complete a study, the resulting information can prove no more accurate than the degree of refinement of approach to data manipulation. Large computers possess no mystic powers to insure accuracy and applicability of results. On the contrary, the capacity for imposing answers upon data increases as the complexity of programming and data manipulation increases.

Data Accumulation

An initial consideration in acquisition of dynamic radionuclide data is definition of the time and space increment pertinent to the physiologic process studied. Some instrumentation permits recording

TABLE II
COMPUTER MANIPULATION OF DYNAMIC
RADIONUCLIDE DATA

Basic Data Processing

1. Digital counts recorded in discrete time and space units.
2. Data labelled as unique studies with parameters of study defined.
3. Correction for instrument dead-time loss and variation in detector efficiency.
4. Editing of data for permanent storage.

Data Manipulation

1. Response of each detector unit delineated by basic curve parameters.
2. Grouping of detector regions with similar responses.
3. Analysis of data from anatomical regions.

Data Display

1. Two-dimensional graph of radioactivity in selected regions.
2. Three-dimensional graph of response over entire detector.
3. Numeric values describing physiologic functions.

of each unique event with positioning coordinates for subsequent interrogation in any number of time and space groupings. Initial reflection might suggest this capacity to hold great benefit for dynamic studies. However, the optimal time and space divisions are determined by anatomic and physiologic characteristics of the organ systems studied and do not vary significantly with individual patients. In general, more than ten observations are required to define a dynamic function. If the mean transit time through the heart is to be studied, an eight second normal mean transit time would indicate that individual time increments be no longer than 0.5 seconds to provide more than 10 observations for the study. Variations from normal which might shorten the mean transit time in patients such as left to right shunting, would suggest that more accurate studies would result from 0.2 seconds accumulation intervals. In contrast, definition of changes in cardiac chambers within each 0.8 second cardiac cycle would indicate need for a concomitant ten fold decrease in count accumulation periods. Description of the spatial separation essential for accurate dynamic studies proves more difficult, but the intrinsic resolution of most detecting devices suggest that square areas with dimensions from 0.5 cm. to 1.0 cm. are realistic. Spatial or temporal division of counts in excess of that required for organ definition limits inspection and interpretation of the large mass of resulting data which introduces potential for significant study error. A typical radionuclide angiocardigram study accumulating counts in each square centimeter over the heart at 0.1 second intervals for a 50 second period would provide approximately 75,000 information units which represents an overwhelming bulk of numbers for physician interpretation. Future development of direct computer analysis of dynamic radionuclide data may significantly reduce this practical limitation on the time and space division of observed radioactivity. However with present methods for data analysis, the least time and space separation of total observed counts that retains adequate resolution for the physiologic function studied supplies the greatest overall accuracy.

Proper labeling facilitates recall of data which is to be permanently stored in magnetic form. Some commercial systems include hardware designed for this purpose, and other systems rely on a software input at some point in data accumulation or transfer. In addition to a unique experiment number, descriptive study parameters can greatly aid later data selection and grouping. Descriptive parameters which have proved useful in our studies include: (1) the number of accumulation periods within a study, (2) the duration of each accumulation period, (3) the subject (patient, phantom, dog, etc.), (4) the organ system, (5) the radiopharmaceutical, and (6) catalog of all corrections performed. Initial time invested in defining all parameters anticipated to be pertinent to the study proves invaluable in later data processing. Another capability which conserves time and expense in permanent data storage on magnetic tape or disk is data editing to remove invalid studies or accumulation intervals of a study which contain needless information. Editing programs should also provide for recovering the remaining portion of a study marred by technical factors such as a broken or scratched tape.

Data Correction

Two imperative corrections for accurate dynamic radionuclide data are for detector field nonuniformity and for instrument dead-time

loss. All stationary detectors demonstrate a degree of non-random variation in efficiency over the field of detection.⁹ The relative counting efficiency at each information point can be determined by comparison of counts observed during exposure to a uniform radiation source. The relative efficiency over the detector must be defined at each range of gamma energy used, and determinations should be repeated at least monthly to assess changes in efficiency. The efficiency ratio is determined for each information point by dividing the average crystal count by the actual crystal count during observation of the uniform source. Use of the average count of all information points in the correction factor permits correction of data without alteration of total observed count. Division of observed counts at each information point by the relative efficiency ratio normalizes data for field nonuniformity.

In addition to limitation of total observed count, instrument dead-time introduces distortion with time in dynamic radionuclide data. The influence of failure to correct for dead-time distortion on data in an excised dog lung preparation is illustrated in Figure 2 which relates the minute volume to the slope of Xe-133 disappearance. Analysis of the same data with and without dead-time correction defines the different relationship as indicated by the different slope, and the greater variation in the relationship as indicated by the different correlation coefficient in uncorrected data.

A number of approaches are available for instrument dead-time determination. One technic individually counts sources with insufficient radiation to influence dead-time and compares the calculated sum to observed counts from the same sources grouped simultaneously over the detector. In another approach, the observed decay of an intense source of a short-lived radionuclide is compared to the true count rate obtained by extrapolation of counting rates after significant decay using the known radionuclide half-life. A variation of these technics which has proved valuable in our experience exposes a small portion of a detector field to a source with a count rate insufficient to be influenced by dead-time losses. Without changing radiation exposed to this region, the total matrix count is gradually increased by bringing an intense radiation source over another portion of the detector field. Typical data obtained with the Digital Autofluoroscope illustrates that count rate falls in a given crystal as the total matrix count increases, and this relationship appears approximately linear below counting rates of 20,000 per second (Fig. 3). The detector saturates at count rates of 25,000 per second and increased radioactivity actually decreases total counts due to anticoincidence circuitry activation. Extrapolation of a linear fit to observed data below matrix counts of 20,000 determines the theoretic maximum matrix count per second at the intercept. This value divided into one second provides the functional resolution time which has been determined to average 24 microseconds for all detector units of the Autofluoroscope. Determination of resolving time by this approach permits dead-time correction by the equation:

$$\text{True counts in each crystal} = \frac{\text{observed count in each crystal}}{1 - \text{resolving time} \times \text{total matrix count}}$$

Minimal computer programming permits this correction for curve distortion which is characterized by depression of curve peaks because of proportionally greater count losses at times of greatest radioactivity (Fig. 4).

Figure 2. Data obtained in an excised lung preparation comparing minute volume with the slope of Xe-133 disappearance are analysed with and without dead-time correction. Failure to dead-time correct defines an incorrect relationship between the two variables and introduces less correlation between the variables. Data distortion caused by dead-time losses at high counting rates must be corrected to assure accurate physiologic studies.

Figure 3. This experiment defines event resolution time for the Autofluoroscope. Insignificant dead-time loss occurs with exposure of an individual crystal (B 7) to a radiation source which produces about 350 counts per second. Without alteration of radiation exposed to this crystal, an intense radiation source is brought over another portion of the detector field to gradually increase the total matrix count. As the matrix counts increase, counts in the individual crystal decline linearly due to dead-time losses. Anticoincidence circuitry introduces nonlinearity as the detector saturation counting rate of 25,000 counts per second is approached. Extrapolation of the linear portion of the data suggests that a theoretic maximum count rate of about 42,000 counts per second or an event resolving time of 24 microseconds best describes instrument performance at count rates less than 20,000 per second.

Figure 4. A standard curve was artificially generated in a localized region of the autofluoroscope detector. The generation process was repeated while a second isolated detector region was exposed to an intense radiation source to increase the total matrix count. Depression of curve peaks characterizes the distortion resulting from dead-time loss. Computer correction of the data using the standard dead-time equation is illustrated in the lower graph.

Other corrections such as application of statistical filters or mathematical correction for collimator depth response may increase the accuracy of dynamic data. However, the validity of these techniques and their usefulness for data enhancement appears severely limited by the low counting rate of present dynamic studies.

Data Manipulation and Display

Proper assignment of observed counts to the anatomic site of origin represents an important determinate of the accuracy of all dynamic radionuclide data. Details of an approach using both the time and intensity of observed radioactivity to provide spatial orientation is included in a separate contribution to this symposium. The potential error in pulmonary function measurement which might result from counts within the left subclavian vein following left arm vein injection of Xe-133 saline in a patient with a left upper lung infiltrate appears obvious (Fig. 5). Careful attention to images of count intensity avoids most errors of quantitation arising from inaccurate anatomic assignment of counts. However, the magnitude of data alteration which may result from minor errors in assignment of areas can prove quite great. Comparison of radionuclide angiocardigram data from correctly assigned cardiac areas with data obtained by moving the same areas only 1 cm. to the left demonstrates severe curve distortion (Fig. 6). All dynamic radionuclide determinations of regional organ function which cannot be oriented to anatomic structure by a consistent, reproducible method must be considered inaccurate. Even with optimal area selection some degree of count overlap from adjacent structures cannot be avoided (Fig. 7).

The final step in data manipulation is the development of time-intensity curves for each detector region of interest. Description of observed data in accepted ways permits characterization of the physiologic function studied. A number of techniques for organizing, correlating and imaging the resulting data aid in appreciation of observed results and are necessary for communication of results among those responsible for patient care. The regrettable ease with which large masses of digital data may be massaged into physiologic answers previously expected to fit the clinical situation demands that each investigator rigorously examine each step of data manipulation in a critical, objective manner. Standardization of techniques and objectively controlled interpretation are factors certain to enhance the accuracy and to facilitate ultimate clinical applications of dynamic radionuclide studies.

Summary

Over the past several years, use of stationary detectors interfaced with a computer for data storage and manipulation have generated widespread interest in dynamic radionuclide studies. A major limitation for these studies is the long event resolution time inherent in all current instrumentation that limits the total count rate to about 25,000 counts per second. In spite of this limitation, a number of studies have suggested that the innocuous nature of tracer techniques for dynamic studies hold great potential benefit for patient management. A general approach to described which is applicable to all dynamic isotope studies. An initial step assigns digital counts to discrete time and space intervals. Data is corrected for distortion from instrument dead-time loss

Figure 5. These computer-constructed three dimensional images at intervals following injection of 30 millicuries of Xe-133 saline into the left arm vein of a patient with a left upper lobe infiltrate demonstrates residual counts in the left subclavian vein near the area of infiltrate. Careful selection of anatomic regions is essential for accurate dynamic radionuclide studies.

Figure 6. The influence of anatomic region selection on the accuracy of dynamic studies is illustrated by this radionuclide angiocardigram data. Only a one centimeter movement of the properly selected atrial regions greatly alters the curves produced.

Figure 7. Data from detector groups at different lateral positions over the precordium illustrate the influence of anatomic configuration on observed data. A certain degree of curve overlap remains even with optimal area selection.

and for variation in detector efficiency over the field of view. Detector regions which correspond to anatomic structure must be accurately delineated, and characterizations of the resulting data which reflect regional organ function are determined.

References

1. D.W. Brown and D.S. Groome, "The Role of the Digital Computer in Nuclear Medicine," JAMA, 203:153-159, (1968).
2. W.J. MacIntyre, J.E. Christie and G. Curtis, "Three-Dimensional Computer Read-Out of Radioisotope Scan Data," Radiology 90:22-26, (1968).
3. R. H. Jones, J.K. Goodrich and D.C. Sabiston, Jr., "Radioactive Lung Scanning in the Diagnosis and Management of Pulmonary Disorder," J Thorac Cardiovasc Surg, 54:520-534 (1967).
4. C.C. Harris, R.H. Jones, T.S. Buffaloe and J.K. Goodrich, "Stationary-Detector Dead-Time Effects on Dynamic Radionuclide Biologic Measurements," J Nucl Med 11:324 (1970).
5. G.L. Brownell, C.A. Burnham, S.M. Pizer, and S. Wilensky, "Numedics: A Computer System for Processing Radioisotope Data from Multiple Sources." Proceedings of Symposium on Sharing of Computer Programs and Technology in Nuclear Medicine pp. 51-60, AEC, Oak Ridge, 1971.
6. J.R. Cox, Jr. and R.L. Hill, III "Interface Design Considerations," Computer Processing of Dynamic Images from an Anger Scintillation Camera Proceeding of a Workshop at Washington University, St. Louis, 1971 (To be published)
7. N.A. Lassen, "Intra-arterial Methods for Measurement of Regional Cerebral Blood Flow in Man," Central Nervous System Investigation with Radionuclides, Gilson and Smoak, pp. 194-202, Charles C. Thomas, Springfield, 1971.
8. R. Hill, J. Clifton, T. Gallager and E.J. Potchen, "Regional Cerebral Blood Flow in Man, II Data Acquisition and Analysis," Arch Neurol 20:384-387, (1969).
9. L.M. Morrison, F.P. Bruno and W. Mauderli, "Sources of Gamma Camera Image Inequalities," J Nucl Med 12:785-791, (1971).

ANATOMIC RESOLUTION IN DYNAMIC RADIONUCLIDE STUDIES BY COMPUTER
IDENTIFICATION OF RADIOACTIVITY FLUCTUATION WITH TIME

Robert H. Jones, M.D., Ronnie B. Klaphaak, B.A.,
and David C. Sabiston, Jr., M.D.

ABSTRACT

Meaningful evaluation of dynamic quantitative radionuclide data obtained with stationary detectors requires assignment of observed counts to known anatomic regions. Previous methods which have been used depend primarily on count intensity and often prove quite laborious. In addition, accuracy of anatomic area delineation varies and the lack of a consistent approach for area determination hinders the clinical usefulness of dynamic radionuclide studies. Two computer programs are described which permit accurate, reproducible descriptions of anatomic areas. Matrix program describes radioactivity curves from each information point of an entire detector field by the parameters: (1) total count, (2) maximum count, (3) time of maximum count, (4) semilogarithmic slope of the curve downsweep, and (5) standard error of the semilogarithmic fit. Comparison of these parameters provides rapid appreciation of regional fluctuations in radioactivity which frequently proves sufficient for relating detector units with similar responses into groups corresponding to discrete anatomic areas. Fourier program provides a more accurate description of the time course of radioactivity by separation of physiologically significant components of complex curves. Curve smoothing by Fourier analysis permits computer recognition of the duration of three most important components of each curve. Subsequent extrapolation of the original data based on this time definition provides actual component curve construction and permits data from a single detector unit to be assigned to several anatomic regions. In addition, this approach holds potential for further automation of dynamic data processing by computer delineation of regions which demonstrate similar time-intensity parameters of component curves.

Department of Surgery
Duke University Medical Center
Durham, N. C. 27710

Introduction

Meaningful evaluation of dynamic quantitative radionuclide data obtained by stationary detectors requires anatomic orientation of detected counts. An early approach provided anatomic resolution by relating the location on x-ray of radiopaque markers with a known spatial orientation to the detector surface.¹ Transmission scanning is a more recent innovation which provides images useful for topographic orientation.² However, maintaining a constant position of the subject during the entire time required to obtain a localizing image and to complete the radionuclide study often proves difficult. Also, neither of these methods provides regional anatomic resolution within an organ. Other approaches interrogate dynamic quantitative data using pictorial displays of count densities to relate detector regions to patient anatomy.³ The comparison of several sequential images required for analysis of patterns of changing radioactivity proves quite time consuming. In addition, low count rates imposed by long count resolution times of current instrumentation hinder accurate delineation of organ regions by count intensity alone. This report describes a method for anatomic resolution based on computer characterization of the time fluctuation of radioactivity in small areas over the detector field. Groups of detector units with similar time responses during a dynamic radionuclide study can be recognized to correspond to discrete anatomic regions. This approach provides more rapid and accurate analysis of dynamic radionuclide data than visual interrogation of pictorial data.

Methods

The method described in this report could be applied to all dynamic quantitative computer data obtained from camera-type detectors. Radionuclide angiocardiodiagrams represent the most common clinical dynamic isotope study, and the geometric configuration of the heart stresses any technic devised to separate adjacent anatomic regions. Therefore, radionuclide angiocardiodiagram data were used exclusively for development and testing of computer programs for identification of anatomic regions. The Digital Autofluoroscope provided all data used for this study. The detector of this instrument which is composed of 294 one centimeter square scintillation probes was positioned over the precordium of supine patients, and 10 mCi of Tc-99m pertechnetate in small volume was injected intravenously. Counts observed by each detector unit at 0.2 sec. accumulation intervals were placed onto computer tape for a one minute period. Data recorded on magnetic computer tape were processed by an IBM 370/165 computer with programs requiring a 300,000 byte memory. The initial data comprised of about 200 records containing spatially oriented counts for each 0.2 sec. accumulation interval were labeled by punch card as a unique experiment. The efficiency of each of the 294 detector units was determined by relative counts obtained during several minute observation of a uniform radiation source, and data were corrected to provide detector uniformity without changing the total matrix count observed at each accumulation interval. A standard dead-time correction was performed using the 24 microsecond Autofluoroscope resolving time determined by separate investigation.⁴ Corrected radionuclide angiocardiodiagram data were stored on disk for subsequent analysis.

Description of Computer Programs

Practical limitations determine the design of computer programs for definition of changes in radioactivity over small regions of the detector field. The most complete description would include digital counts and isotope intensity curves for each data point for the entire study duration. However, the computer expense and analysis time required for this extensive data manipulation would not prove practical for routine clinical use. Delineation of important curve characteristics such as the peak count, time of peak count, total count, and the down-slope of the curve provides an alternate approach for curve description. A numeric output of these parameters in a spatially oriented matrix might allow rapid appreciation of the time course of radioactivity over each probe. Adjacent detector units which demonstrate similar time responses could be grouped into larger areas that relate to anatomic structure.

The two computer programs developed to describe basic curve parameters were titled Matrix and Fourier.⁵ Matrix program provides the maximum, time of maximum, area, semilogarithmic down-slope, and the standard error of the semilogarithmic fit for each curve from the 294 data points. Data card input selects the proper study and the records within the study desired for interrogation from permanent disk storage. The entire array of original matrix data is read into a buffer memory, and the maximum count, time of maximum count, and sum of all counts for the study duration are identified for each matrix point. An IBM scientific subroutine for least squares analysis fits a line to the semilogarithmic data from the point of curve maximum to study termination and calculates the slope and standard error of the fit. Matrix program output supplies the numeric value of all five parameters for the entire detector matrix which is stored on disk and printed on paper.

The Fourier program provides a more accurate time description of curves from individual probes to further aid anatomic resolution and to permit separation of component parts of complex curves. The input of Fourier program is the same as for Matrix program. Data from individual probes must be smoothed to permit computer recognition of true curve fluctuations from random statistical variation. Examination of a variety of smoothing technics by comparison of smoothed curves to original data identified Fourier analysis as the approach which removed the greatest amount of random fluctuation with the least curve distortion. Fourier analysis provided optimal smoothing of the wide range of complex curves which may be recorded from small regions over the precordium during tracer passage through the heart. A Fourier series sums a number of harmonic components which construct a curve to describe the original data.

Fourier analysis is illustrated using data from a single detector which suggests a curve with one primary component and recirculation (Fig.1). Only 110 counts are observed during the 0.2 sec. accumulation interval with greatest radioactivity, and marked statistical fluctuation present in the original data reflects the low count rate. Curves of the second through the fifth frequencies of a Fourier series illustrate that each subsequent frequency enhances curve definition but adds less total information. An optimal number of frequencies would provide an adequately smoothed

Figure 1. Statistical fluctuation in counts recorded by a single probe during 0.2 second intervals of a radionuclide angiogram illustrates the need for data smoothing. Curves of frequencies 2 through 5 of the Fourier series demonstrate that each successive frequency enhances curve resolution but contains less total information. The curve produced by summation of the first five frequencies fails to adequately describe the observed data. The sum of 20 frequencies of a Fourier series provides too close an approximation of the original curve and retains excessive random fluctuation. Ideal smoothing of this data results from summation of 10 frequencies of a Fourier series.

curve which retained all major inflections. In this data, the first five frequencies appear inadequate for accurate curve description. The sum of twenty frequencies reflects every major fluctuation in the original data and this faithfulness of curve reproduction hinders computer recognition of curve components. The optimal smoothing which inflicts least curve distortion for this data appears to be obtained by the sum of 10 frequencies. More complex data arises from detector units over organ borders such as the right heart and lung (Fig. 2). The observed curve appears to include data from tracer passage through the right heart, lung and recirculation of radionuclide through the vena cava. Fourier smoothing using 10 frequencies fails to separate right heart counts from radioactivity in the lung. The curve constructed from 15 frequencies of a Fourier series separates counts in the right heart, lung and vena cava. With 20 frequencies, a second maximum appears in the curve component containing lung counts which probably results from organ movement with respiration. This second peak in data from a single anatomic site reflects inadequate smoothing and would hinder accurate computer separation of physiologic components of the curve. In data from this detector, 15 frequencies of a Fourier series produces an optimally smoothed curve containing all fluctuations of physiologic importance.

Variation in the types of curves recorded from discrete regions over the precordium demands construction of each optimally smoothed curve by a different number of frequencies of a Fourier series. Using the IBM subroutine for Fourier analysis, a modulus describes the magnitude of contribution of each Fourier frequency to the final constructed curve. Empiric observations defined modulus limits which included the number of frequencies that provided optimal smoothing. For each detector unit, the modulus of each frequency was compared to established criteria to select the number of frequencies that optimally smooth the data. Smoothed data from each probe are stored in a memory without destruction of original data.

The second step of Fourier program uses smoothed data to define the duration of important curve components. Each inflection point of the smoothed curve was identified as a maximum or minimum, and time and intensity values for each maxima and minima were stored in memory. Observation of angiocardigram data and consideration of the geometric configuration of the heart suggested that three curve components would describe all significant biologic events observed by any single detector. Minimum points adjacent to the maximum with greatest count intensity identified the duration of the primary curve component, and time the primary curve component was stored in memory. Criteria which best separated the second and third components were obtained by repeated observations on a variety of angiocardigram data. The maximum with second greatest count intensity was identified and any maxima adjacent to this point with counts within 20% of the peak value were included in the secondary component. The minimum immediately preceeding and following the selected group of maxima defined the duration of the second component. The largest maximum not included in the primary or secondary component curves was identified and all adjacent maxima within 30% of the peak value were included in the tertiary component. To insure that the duration of curve components be exclusive, the time limits of previously determined components limited the duration of subsequent components. The Fourier program provides an optional paper print-out of the matrix arrays in memory which defines the time and intensity values of beginning, peak and end of the three component curves for each of the 294 data points.

Figure 2. Radionuclide angiogram data from a single detector situated over the right heart-lung border illustrates that optimum smoothing must also retain all significant physiologic information. The first ten frequencies of the Fourier series fails to separate right heart and lung counts. The sum of 15 frequencies provides ideal smoothing, whereas the sum of 20 frequencies introduces a second peak into the lung data caused by respiration. The second peak would be recognized incorrectly by the computer as separate physiologic information.

The final step in Fourier program constructs actual component curves from the original data based on the component duration identified on smoothed data. The biologic functions examined with radionuclides are assumed to follow a semilogarithmic course. Overlapping areas of component curves are assumed to best be separated by extrapolation of the component with greatest counts. Component curves were constructed from original data to minimize data distortion introduced by curve smoothing. A semilogarithmic curve is fitted by least squares analysis to the upsweep and downsweep of all original data points within the duration of the primary component defined upon smoothed data. Extrapolation of the semilogarithmic function determined provides the origin and completion of the primary component. The complete primary curve is subtracted from the original data and the remaining counts are retained in memory for further interrogation. Secondary and tertiary curve components are sequentially separated by semilogarithmic extrapolation except that the extrapolated data is retained only when the magnitude of these counts does not exceed the observed counts which remain in the original data. Counts of observed data which remain after removal of the three component curves describe a fourth curve to insure that no significant amount of detected radioactivity will be ignored in study interpretation.

Program Application

The Matrix program provides analysis of dynamic radionuclide data to permit rapid appreciation of changes in radioactivity within each information point over the detector field. Radionuclide curve parameters such as peak count, time of peak count, total count, and logarithmic slope of radionuclide disappearance permit initial data evaluation. The program could also be easily modified to calculate other mathematical descriptions of isotope curves which might provide a more accurate description of specific physiologic functions. The usefulness of individual curve parameters calculated by the Matrix program depends primarily upon the time course of the physiologic function studied. Slow biologic functions permit analysis by description of count intensity, whereas anatomic definition in organs with more rapid physiologic function proves more accurate when defined by the time course of the curve. Data from a Tc-99m pertechnetate perfusion study in excised dog lungs illustrated the use of Matrix program for anatomic resolution (Fig. 3). Motion of respiration and irregular lung geometry hinders definition of lung borders by visual inspection of the preparation. The Matrix program provides rapid initial data analysis by calculation of the sum, peak, and time of peak of each detector unit. The total counts during perfusion provides useful demarcation of lung borders. In general, all detector units with counts greater than 1000 appeared to record biologic data and background counts averaged about 200. Inspection of total counts also insures that no region with a significant amount of radioactivity is excluded from data analysis. However, attention to total counts alone does not separate counts within the perfusion tubing from counts which originate in the lung apex. The maximum count observed by each detector unit also delineates lung borders, and all peak counts greater than 100 appear to represent lung counts with maximum background counts of about 20. High peak counts observed along the inflow perfusion tubing prohibit accurate spatial resolution in this region. The time of maximum count observed by each crystal aids separation of detector regions at the lung apex. Each unit

Right Lung
Flow
1220cc/min

Left Lung
Flow
325cc/min

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20
A	221	614	128	381	563	908	908	1500	940	1750	1750	3000	1900	4200	1000	1000	1000	1000	1000	1000
B	342	505	328	602	888	2357	7000	2400	2662	7000	6002	5000	4902	5330	4300	3000	2177	000	131	305
C	136	237	439	1705	3759	3904	3369	5071	0400	9500	6000	5000	6006	0900	5430	4762	3766	3231	320	163
D	240	419	1360	1176	6360	3099	680	7700	7300	7316	7616	7670	9009	0000	700	5300	6700	2413	000	160
E	776	1076	1007	1011	7170	7306	0900	10707	10710	9500	11250	9067	0199	0000	5000	1900	1750	2500	906	450
F	600	1310	2104	4100	1215	5217	5604	0017	10000	3612	3000	3002	3236	5015	0916	4200	2000	1000	600	100
G	1409	5051	3370	6154	4150	4755	4350	4650	9210	0007	0000	0002	0000	0000	3000	2007	2000	1000	1000	170
H	112	2700	2700	2610	5110	1675	0000	3090	0000	0900	1077	3000	1900	3053	4300	4767	1000	1007	370	206
I	2470	3406	3406	0561	1700	0600	0400	6204	0000	7002	0009	7115	0275	0670	5637	4700	4700	1705	366	100
J	136	739	1307	2709	5070	0017	0000	0009	0610	0277	0000	0006	0995	0000	6002	5000	3400	2400	1002	210
K	1401	1167	1000	1265	0170	5002	5000	6000	0600	5011	0700	0600	7072	0600	4753	3000	2901	1000	071	160
L	275	200	740	169	090	0625	1001	1112	1067	3500	0211	0007	6600	0301	3490	2662	1000	700	107	00
M	310	150	110	062	090	900	100	031	437	2070	1900	1007	1750	1772	1007	1000	100	170	100	100
N	423	160	150	200	210	200	100	110	900	000	000	000	012	711	000	213	063	019	00	000

C

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20
A	3	5	7	12	8	9	7	6	11	10	7	11	6	9	9	9	9	6	4	4
B	9	6	5	8	6	7	6	6	8	6	6	5	5	5	5	5	5	5	5	5
C	6	7	5	6	6	8	7	4	6	6	6	5	9	5	7	7	9	9	10	10
D	11	12	11	6	5	8	6	5	4	4	4	5	5	6	7	7	12	9	4	4
E	3	4	7	5	7	6	6	4	6	5	4	6	6	6	6	7	7	7	5	5
F	3	4	3	3	4	5	5	6	5	4	4	5	4	5	7	6	6	7	6	6
G	3	3	3	1	3	4	4	4	4	5	5	5	5	6	6	6	6	6	6	6
H	7	7	8	6	6	6	8	6	6	6	6	8	5	7	7	10	6	8	7	6
I	7	5	4	6	6	6	7	7	6	5	5	6	7	7	8	8	8	6	6	6
J	6	6	6	7	8	8	9	10	6	6	5	5	5	5	6	9	6	7	7	6
K	8	41	37	37	33	20	10	11	7	9	6	7	5	6	7	8	8	7	7	7
L	8	7	6	60	7	29	12	11	10	7	8	6	7	6	9	8	10	8	10	13
M	6	8	17	10	24	11	9	2	9	8	7	8	6	6	6	5	6	5	5	5
N	7	7	6	9	6	6	6	9	6	11	9	9	9	9	7	6	12	7	11	5

D

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20
A	24	50	24	32	35	20	13	111	94	123	102	90	100	160	61	72	69	39	17	9
B	22	25	45	65	78	106	154	232	218	547	277	302	316	419	295	266	163	54	33	18
C	39	20	43	100	213	220	220	389	342	410	527	435	053	686	425	329	235	157	81	10
D	19	40	85	317	566	470	413	732	510	641	972	706	003	755	600	410	320	137	54	14
E	125	105	205	628	642	533	727	1064	1041	904	1109	1063	069	684	426	561	211	163	55	21
F	110	177	147	261	283	370	532	917	1034	1205	1200	1073	605	599	300	159	172	93	50	25
G	170	415	679	719	626	649	917	650	1005	0006	0900	072	920	373	270	100	134	90	49	29
H	332	331	272	332	300	412	329	390	005	507	046	007	062	194	049	90	90	09	09	10
I	353	362	435	609	501	439	660	607	695	547	007	707	000	250	335	477	100	120	57	10
J	57	59	60	69	109	215	206	007	047	062	990	1053	1090	009	030	530	323	007	123	17
K	34	72	64	133	170	201	270	205	300	439	671	606	717	503	300	104	237	161	61	19
L	21	23	19	34	66	64	87	84	100	331	054	050	309	512	335	204	73	63	30	18
M	10	10	17	23	20	31	41	47	20	106	176	105	132	173	09	109	20	30	14	13
N	24	24	46	30	17	13	28	32	26	66	01	50	33	66	90	36	29	12	17	26

Figure 3. Data from an excised dog lung perfusion study illustrates the usefulness of Matrix program for definition of anatomic borders. A. Excised dog lungs with each main stem bronchus and main pulmonary artery individually cannulated are positioned over the detector. B. Total counts over the period of perfusion aid in delineation of lung borders. C. The time of maximum count demonstrates high counts in row G to occur before the maximum counts in the lung hilum thereby confirming the origin of these counts to be from perfusion tubing. Late counts in the left upper lung reflect the outflow of blood from the left atrial cannula. D. The maximum count observed by each detector unit during period of study further aids definition of lung borders.

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21			
A	25	24	22	21	24	18	24	21	19	25	24	20	19	21	22	24	23	26	25	142	120			
B	5	17	21	20	22	27	17	R	20	21	21	23	17	21	22	22	22	22	26	133	146	131		
C	5	5	21	20	21	19	21	P	21	19	24	27	26	23	24	19	6	20	27	122	117	131		
D	5	5	5	24	26	30	13	A	37	20	23	5	5	R	6	6	6	6	6	98	131	134		
E	5	5	5	5	5	5	44	44	37	41	38	7	A	6	5	6	6	6	6	146	124	129		
F	5	5	5	5	5	47	45	44	44	37	41	8	7	5	6	6	R	6	7	134	112	139		
G	41	42	46	46	41	44	41	33	41	34	29	8	8	7	6	6	V	8	8	12	126	104		
H	48	48	50	A	48	49	11	13	12	12	12	38	P	8	7	7	7	7	8	8	8	117	141	
I	24	47	52	O	54	45	10	10	12	10	11	34	A	31	9	10	10	8	7	8	12	53	69	
J	22	23	21	51	10	13	13	10	10	13	9	39	39	36	39	35	51	48	11	61	56			
K	18	20	24	24	18	23	23	23	45	38	49	35	35	35	39	47	L	43	39	59	56	54		
L	22	24	20	18	22	25	27	L	22	34	25	30	L	36	39	31	35	51	V	39	43	51	60	118
M	21	25	22	21	23	23	24	A	24	23	31	27	A	31	31	39	39	43	51	47	107	60	66	
N	25	27	27	19	22	19	15	26	23	29	22	23	21	26	31	27	40	60	140	89	67			

Figure 4. Data from a typical radionuclide angiocardigram illustrates spatial orientation obtained by time resolution. Each time unit represents 0.2 seconds of accumulated data and 0.23 seconds of real time. The timing of bolus passage through the central circulation provides discrete anatomic resolution which permits grouping of detector units corresponding to anatomic structure.

	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21
A	39	39	67	73	75	76	73	99	84	101	86	96	121	92	111	83	87	78	48	60	60
B	58	44	60	75	88	96	96	115	109	128	85	83	112	102	105	99	85	68	65	64	49
C	110	80	68	68	71	98	113	166	103	127	108	99	95	86	91	96	60	58	63	64	48
D	175	70	46	68	85	115	98	121	117	118	149	145	109	209	129	221	79	62	49	51	60
E	239	203	133	120	176	165	103	149	158	149	130	193	186	306	256	319	232	59	45	48	45
F	70	141	111	137	143	174	194	191	152	165	185	148	233	432	482	558	347	101	53	43	50
G	48	73	123	102	138	133	125	194	188	224	158	179	333	407	470	453	391	156	74	49	62
H	58	77	151	142	132	132	153	251	188	215	193	203	268	333	321	374	300	246	111	54	54
I	74	76	84	112	98	198	193	312	270	203	200	162	188	166	197	214	210	176	117	62	58
J	53	79	82	93	95	127	133	240	215	174	202	175	155	120	124	139	138	101	74	63	77
K	58	58	68	92	99	115	108	118	104	134	146	154	162	160	173	119	136	102	69	67	57
L	96	58	72	90	93	107	119	131	119	118	87	133	110	125	136	140	152	127	55	50	60
M	89	59	89	90	96	112	100	124	89	104	91	119	108	117	90	128	93	86	60	59	78
N	49	52	58	56	63	93	74	95	63	92	73	105	99	105	49	74	93	45	52	56	68

Figure 5. The intensity of peak counts for each detector unit during the radionuclide angiocardigram in Figure 4 suggests the validity of region definition. Sequential decrease in maximum count as the radionuclide progresses from the right ventricle through the central circulation reflects gradual dilution of the tracer bolus. However, definition of cardiac regions by count intensity alone would prove difficult.

Figure 6. These radionuclide angiogram curves obtained further confirm the validity of anatomic regions selected in Figure 4.

Figure 7. Images of the total count and maximum count during study calculated by Matrix program reflect regional perfusion in excised dog lungs.

Figure 8. Images of slopes of the semilogarithmic disappearance of radioactivity calculated by Matrix program are particularly useful to depict regional air exchange in the lungs. The left image illustrates regional ventilation in a normal six year old child and the right image depicts air exchange rates in the lungs of an asthmatic child during an asymptomatic period.

of time represents 0.5 seconds, and the radionuclide bolus peak appears to arrive in the right lung hilum at 2 seconds and at 2.5 seconds in left lung hilum. Early peak counts along the inflow perfusion tubing can be recognized to originate from a source other than lung. The time of maximum count demonstrates a gradual increase from lung hilus to lung periphery, and background counts in crystals outside the lung border demonstrate a random time course. Thus, comparison of the sum, peak, and time of peak provided by the Matrix program permits an approximation of radioactivity curves from each detector unit which aids selection of crystals with similar responses for further analysis. The simultaneous description of both time and intensity parameters enhances the accuracy of this approach.

Time parameters describe rapid biologic functions more efficiently than do parameters of count intensity. A radionuclide angiocardioqram obtained by direct injection of Tc-99m pertechnetate into the right ventricle of a patient illustrates the use of time for anatomic resolution (Fig. 4). Each time unit represents 0.2 seconds of accumulated data and 0.23 seconds of real time. Early counts describe the catheter course and permit separation of the right ventricle from the adjacent left ventricle which demonstrates later counts and from the liver which demonstrates extremely late peak counts. The timing of the bolus passage through the central circulation provides discrete anatomic resolution. The intensity of peak counts at each of these times demonstrates a sequential decrease in maximum count reflecting gradual dilution of the radionuclide bolus as the tracer progresses from the right ventricle through the central circulation (Fig. 5). The configuration of radionuclide curves obtained from regions identified by the time of peak count further confirms the accuracy of anatomic resolution by this approach (Fig. 6). Definition of cardiac regions by the count intensity alone would prove difficult and could be done only by sequential interrogation of individual accumulation periods. This commonly used approach relies primarily on time differences in count intensity, but selection of cardiac regions proves most accurate and rapid when the regional time course of radioactivity is thoroughly identified by computer analysis.

Disk storage of data calculated by Matrix program also permits construction of images of the data. Images of selected curve parameters from information points over the entire detector field permits visualization of organ function which cannot be readily attained without computer manipulation.⁶ In addition to pictures of total counts during a dynamic study, peak count over each detector unit may be imaged independantly of time (Fig. 7). Images of slopes of the semilogarithmic disappearance of radioactivity may reflect regional organ function and are particularly useful for description of regional ventilation rates (Fig. 8).

Computer definition of time of peak count provides a useful approach for initial assessment of dynamic radionuclide studies, but this single parameter holds limited accuracy in complex curves of counts which arise from organ borders and adjacent anatomic regions within organs. The Fourier program was developed to achieve a more accurate time description of radioactivity fluctuation and to utilize this time resolution as a basis for separation of the components of complex curves. Data recorded over the cardiac septum includes both right and left heart counts and well illustrates the problem of spatial resolution in organs with complex

geometric configurations. Computer separation of individual components of complex curves might enhance recognition of detector units with similar time-intensity curves and also permit grouping of component curve data from any single detector unit into several different regions corresponding to anatomic structure. Using this approach, early counts from a detector situated over the intraventricular septum would be recognized to arise from the right ventricle whereas later counts would be assigned to the left ventricle.

Radionuclide angiocardioqram data in a young child with a ventricular septal defect demonstrates typical complex curves observed by probes over the intraventricular septum (Fig. 9). The time of maximum of the primary and secondary component curves determined by Fourier program for each detector unit depicts the flow of radioactivity over the detector field (Fig. 10). Timing of the passage of tracer bolus through the central circulation by the primary component permits recognition of anatomic regions with sufficient definition to identify the right and left heart chambers and to separate the aortic arch and pulmonary outflow track. The time of maximum of the second component curve for each data point further enhances anatomic resolution. Along the cardiac septum, secondary curve components are identified within the primary right heart region which contain left heart counts, and a zone of detector units within the left heart demonstrate a secondary component with a time of maximum which corresponds to the right heart peak time. In addition, the aortic arch region contains a secondary component of radioactivity arising from the pulmonary outflow track, and the pulmonary outflow track contains secondary counts with a time of maximum suggesting origin from the aortic arch.

In addition to enhancement of anatomic resolution, definition of curve components permits subgrouping of data from a single detector unit which correspond to adjacent anatomic regions. Computer identification of the duration of the primary and secondary components of each curve over the heart in this patient is illustrated in Fig. 11. A semilogarithmic extrapolation of original data beyond the time defined as the duration of curve components provides the origin and completion of each component curve (Fig. 12). Extrapolation from original data proceeds sequentially beginning with the primary curve component. Therefore, constructed curves may be expected to prove most accurate for the primary component and least accurate for tertiary components which contribute the least total information to the study. Construction of curve components is limited to prevent generation of data in excess of actual observed counts. Thus, the three component curves separated when added to the remaining original data which describes a fourth curve equal the observed data. The separation of data observed by a single detector unit into more than one group which correspond to different anatomic regions would appear to enhance the accuracy of quantitative radionuclide angiocardioqram data. In addition, the approach of Fourier program for curve component separation appears well suited for further automation in processing of cardiac transit studies. Work is currently in progress to extend Fourier program to include computer selection and grouping of components of adjacent detector regions with similar time-intensity responses. Computer manipulation of the total data from the detector groups selected could describe the mean transit time, area under the curve, or other important curve parameters describing flow in discrete cardiac areas.

Figure 9. Radionuclide angiogram data in a young patient with a ventricular septal defect illustrate the complex curves which arise from individual detectors which view adjacent anatomic structures. Detector E8 is over the right ventricle, and unit J8 is over the left ventricle. Both unit G8 and H8 overlie the intraventricular septum.

Figure 10 A. The time of maximum count of the primary component from each detector indexes the passage of radioactivity over the detector surface. Each time unit represents 0.23 seconds real time.
 B. The time of maximum of the second component curve further enhances anatomic resolution by delineating secondary counts which arise from adjacent anatomic structures.

Figure 11. The duration of the primary and secondary components of each curve as determined by Fourier program using smoothed data provides the time reference necessary for computer construction of component curves by extrapolation.

Figure 12. Extrapolation of original data permits separation of observed curves into component curves which describe data from different anatomic regions. The component curves are constructed without generation of new data, and the resulting component curves when added equal the observed data.

Summary

Meaningful evaluation of dynamic quantitative radionuclide data obtained with stationary detectors requires assignment of observed counts to known anatomic regions. Previous methods which have been used depend primarily on count intensity and often prove quite laborious. In addition, accuracy of anatomic area delineation varies and the lack of a consistent approach for area determination hinders the clinical usefulness of dynamic radionuclide studies. Two computer programs are described which permit accurate, reproducible descriptions of anatomic areas. Matrix program describes radioactivity curves from each information point of an entire detector field by the parameters: (1) total count, (2) maximum count, (3) time of maximum count, (4) semilogarithmic slope of the curve downsweep, and (5) standard error of the semilogarithmic fit. Comparison of these parameters provides rapid appreciation of regional fluctuations in radioactivity which frequently proves sufficient for relating detector units with similar responses into groups corresponding to discrete anatomic areas. Fourier program provides a more accurate description of the time course of radioactivity by separation of physiologically significant components of complex curves. Curve smoothing by Fourier analysis permits computer recognition of the duration of three most important components of each curve. Subsequent extrapolation of the original data based on this time definition provides actual component curve construction and permits data from a single detector unit to be assigned to several anatomic regions. In addition, this approach holds potential for further automation of dynamic data processing by computer delineation of regions which demonstrate similar time-intensity parameters of component curves.

References

1. R.H. Jones, J.K. Goodrich and D.C. Sabiston, Jr., "Radioactive Lung Scanning in the Diagnosis and Management of Pulmonary Disorder," J Thorac Cardiovasc Surg, 54:520-534 (1967).
2. H.O. Anger and J. McRae, "Transmission Scintiphotography," J Nucl Med, 9:267-269 (1968).
3. Y. Ishii and W.J. MacIntyre, "Measurement of Heart Chamber Volumes by Analysis of Dilution Curves Simultaneously Recorded by Scintillation Camera," Circulation, 44:37-46 (1971)
4. C.C. Harris, R.H. Jones, T.S. Buffaloe and J.K. Goodrich, "Stationary-Detector Dead-Time Effects on Dynamic Radionuclide Biologic Measurements," J Nucl Med, 11:324 (1970)
5. R.H. Jones, R.B. Klaphaak, C.M. Coulam, J.K. Goodrich and D.C. Sabiston, Jr. "Anatomic Resolution in Dynamic Radionuclide Studies by Computer Identification of Radioactivity Fluctuation with Time," J Nucl Med, 12:371 (1971).
6. R.H. Jones, C.M. Coulam, J.K. Goodrich and D.C. Sabiston, Jr., "Radionuclide Quantitation of Lung Function in Patients with Pulmonary Disorders," Surgery 70:891-903, (1971).

**Clinical Applications of Computers -
Nuclear Angiocardiology**

**D. D. Watson, Ph.D.
J. P. Nelson, M.S.
S. Gottlieb, M.D.**

**Mt. Sinai Medical Center
Miami Beach, Florida**

April 1972

Abstract

The application of computers to nuclear angiocardiology has proven exceedingly fruitful. Pre- and post-operative studies in patients undergoing cardiac surgery are being performed with a scintillation camera/computer system to provide relative indices of patient status. A simple mathematical model has been developed to demonstrate the feasibility of extracting such parameters as ejection fraction from activity histograms. The continued development of such procedures should yield non-invasive methods for the evaluation of myocardial function which will be highly beneficial in the diagnosis and treatment of cardiac disease.

Index words: clinical computers, nuclear angiocardiology, dynamic isotope studies

Clinical Applications of Computers - Nuclear Angiocardiology

**D. D. Watson, J. P. Nelson, S. Gottlieb
Mt. Sinai Medical Center
Miami Beach, Florida**

Introduction

In our presentation at the symposium of a year ago, we described the Hewlett-Packard computer based data system which two of us had recently implemented at Mt. Sinai.¹ Since then we have made only minor alterations in the system, basically because our applications oriented philosophy has not changed. Our system facilitates simple use of the basic functions common to nearly all such systems: data acquisition, variable time-frame image formation in a 64 x 64 grid, and curve generation from an area designated by a cursor. We therefore will not further discuss the system, but rather present preliminary data from one of the more exciting of the several applications we have developed and pursued in the past year.

Our interest in the coordination of spatial and temporal information led us quite naturally to the study of cardiac function. The increasing interest in the assessment of cardiac function, the lack of truly non-invasive visualization techniques, and the excellent foundations laid by others in the isotope field, challenged us to pursue such work at our own lab. We have formed what we believe to be significant concepts in the course of our work, and we hope that these methods, when combined with other non-invasive procedures such as EKG, VKG, and diagnostic ultrasound, will enable us to pursue a truly non-invasive multi-disciplinary approach for the study of cardiac function.

The patients selected for this continuing study are those undergoing heart surgery. In consultation with the thoracic surgeons, it was felt that this group presented the greatest opportunity, due in part to the recent interest focused on the assessment of cardiac function following surgery. Since multiple catheterizations are undesirable, some easily obtained relative indices of the patients' status are potentially beneficial. Also, the extensive clinical workup undergone by these patients is a source of a wealth of corroborative information seldom available. Finally, the chance to perform repeated studies is of great importance in determining the limitations of our testing procedures.

Clinical Procedure

The patient is placed supine on a canvas stretcher beneath a scintillation camera which is connected to the computer system. A sheet source of ^{99m}Tc - pertechnetate is placed beneath the patient and the patient positioned to encompass the cardiac silhouette. A transmission view of the cardiopulmonary region is then recorded. After completion, the source is removed and the patient injected as described below with 10-15 mCi ^{99m}Tc - pertechnetate. Serial images are recorded both by the camera and the computer at one second intervals for thirty seconds. The nuclide is allowed to equilibrate for two or three minutes and a single static view is then recorded. This simple

procedure requires less than ten minutes. Should views other than these A-P projections be desired, the protocol may be preceded by a similar injection of a small volume of ^{99m}Tc - sulfur colloid which is subsequently cleared from the blood pool by the liver. A second injection of pertechnetate can then follow in a few minutes.

Since our prime goal is to develop a simple and truly non-invasive technique, we have restricted our isotope delivery to intravenous injections into antecubital sites. This quite naturally requires that certain precautions be taken to ensure rapid bolus introduction into the right heart. The reliable introduction of a single bolus of tracer is mandatory for application of critical diagnostic techniques, so here we will digress briefly to describe the method we employ.

We are able to reliably deliver the nuclide to the right heart by the Oldendorf technique² as a rapid single bolus when the basilic vein or one of its tributaries (Fig. 1) is used as the site of injection. Delivery into the more laterally located cephalic group often results in spreading of the bolus in time and frequently in actual division of the bolus into two discernible components. Best results are obtained when the arm and shoulder are maintained in a neutral position approximate to the patient's normal posture.

Figure 2 is a comparison of two 2 mCi ^{99m}Tc - sulfur colloid injections in the same patient. This sequence demonstrates the rapid bolus component into the axillary vein and the delayed component in the cephalic vein when the cephalic injection site is chosen. A single rapid bolus transit into the axillary vein is observed when the basilic injection site is chosen. The effects of the different injection sites on the resultant count rate histograms is quite profound. The appearance of serial scintiphotos was also markedly altered by the choice of injection site.

Data Review & Analysis

Processing of the resulting voluminous patient data has been largely made possible by judicious use of the computer, allowing us to rapidly and simply review the results and extract pertinent information and enabling the routine performance of these studies with a minimum of required physician time. At the present time, a nuclear angiocardiographic study may be performed and the data reduced to an analyzable form in about thirty minutes.

After completion of the study, a visual review of the data accumulated by the computer is made. From this review cursors are located in the superior vena cava, right ventricle, lung field, and left ventricle. Count rate histograms from these areas are then printed. These are used to determine the integrity of the injection, and initially for optimal interval selection for photos of the right heart, lung perfusion, and left heart phases of bolus transit. Anomalies are also highlighted in this manner by selecting image time and interval to correspond to anomalies in the histograms. In addition, we produce an image encompassing the entire first transit of the bolus through the heart, an equilibrium view, and a transmission view. We often generate a composite picture of first transit, equilibrium and transmission to outline the intra- and extra-cardiac borders. Pericardial effusion may often be more clearly displayed in this fashion (Fig. 3).

We are also able to extract much physiologic information from the data which are stored and presented to us via the computer. We will discuss briefly several techniques by referring to some examples taken from our studies.

Figure 1 A sketch of the superficial veins. The dashed lines indicate the medial injection sites which communicate with the basilic vein and the lateral injection sites which communicate with the cephalic vein.

Figure 2 Two 2 mCi injections of ^{99m}Tc - sulfur colloid given the same patient in the same arm. The upper rows show a basilic injection with single compact bolus entry. The lower rows show a cephalic injection with separation of the bolus into two components.

PERICARDIAL EFFUSION

Figure 3 Pictures taken from the computer CRT showing the superposition of the lung transmission image with the cardiac and liver image. The bottom row are different time intervals chosen to display right heart, lung perfusion, left atrial and ventricular filling, and left ventricle, respectively.

Figure 4 shows a typical series of the histograms which are printed on every cardiac study. The presence of a very small atrial septal defect caused the recirculation peak noted in the figure; the shunt flow in this case being only 10%. Such measurements may be estimated either by: a) comparing the area of the lung circulation peak to the area of the lung recirculation peak; or b) by using the conventional indicator-dilution technique to measure the cardiac output separately at the left and right ventricles. The former method works to the extent that the recirculation peak can be defined. The latter method results in a direct measurement of pulmonary and systemic cardiac flow but has not been found reliable due to errors that occur in the measurement of flow from an individual cardiac chamber. These errors arise from incorrect compensation for background activity originating outside the cardiac chamber in question.

The use of the C2/C1 ratio of the lung curve for shunt detection has been explored here and elsewhere. (C2/C1 is the ratio of the curve height measured at twice the buildup time to the height at the curve peak.) The C2/C1 ratio works well for shunts which are large enough so that individual recirculation peaks are not discernible. In such cases the lung curve appears to have rapid buildup and prolonged washout. It must be pointed out however, that an elevated C2/C1 is not specific to an intracardiac shunt since various valvular lesions and myocardial dysfunctions may also give rise to elevated C2/C1 ratios. The C2/C1 ratio may also fail in the presence of very abrupt bolus injections such as those we achieve in our lab. However, a prominent shunt recirculation peak will be present in these cases.

Cardiac output can be determined from the flow curves by measuring C_f/A where A is the area under the count rate curve, excluding recirculation; and C_f is the count rate after equilibration of tracer concentration. Our main difficulty with this technique has been that in theory C_f should include only intracardiac activity. In fact extracardiac activity is a significant portion of the externally monitored count rate. Attempts to subtract the extracardiac activity meet with variable degrees of success when one attempts to record counts over a single cardiac chamber or great vessel. We find that more consistent results are obtained by counting activity over the entire heart and integrating both left and right heart peaks together as had been done with the precordial probe. This technique is illustrated in Fig. 5. It should be noted that our cardiac output measurements are done in the AP projection to eliminate interference from lung activity.

For pre- and post-operative assessment of cardiac function, we have found useful the mean transit times from right ventricle to lung and from right to left ventricle. The value of using the mean times is, of course, that it directly relates flow and volume between cursor locations.

$$\bar{t}_{ab} = \frac{V_{ab}}{F} \quad \text{where a and b denote cursor locations}$$

When combined with the cardiac output, these measurements will provide indications of alterations in central blood flow and/or volume.

The count rate histograms often reflect striking improvement in the patient's condition following surgery. Figure 6 illustrates comparative right atrial curves from a patient with severe mitral insufficiency. After replacement of the faulty valve the pulsatile structure disappears, consistent with the report of relative tricuspid insufficiency. Figure 7 illustrates the remarkable shortening of transit time and improvement in bolus structure in a patient who underwent aortic valve replacement.

Figure 4   Histograms obtained from a cardiac flow study. Complete entry of the bolus into the right heart occurred in less than one second.

Figure 5 Histogram formed from a cursor located over the entire heart. Better statistical definition and more consistent ratios of intracardiac to background activity can provide higher accuracy for cardiac output determinations in this way.

Figure 6 Histograms formed from a cursor over right atrium. The large pulsatile structure in the pre-op study prior to a mitral valve replacement was taken to be due to reported relative tricuspid insufficiency.

Figure 7 Comparison of LV curves from the same patient pre-op and post-op following an aortic valve replacement.

The search for a more direct indicator of myocardial function led us to attempt to correlate the C2/C1 ratios from the right ventricle (RV) and left ventricle (LV) curves with ventricular ejection fractions. The C2/C1 ratios were found to be overly affected by factors other than the ejection fractions to be of much value for that purpose. To help understand our problem and guide us to a more fruitful approach we developed a very simple mathematical heart-lung model. The right and left heart are represented by two pumps with constant ejection fractions. The bolus transit through lung and extra-cardiac vasculature is assumed to follow a gaussian distribution. The important outcome of this model is the description of the activity curve from the left ventricle. The functional form is:

$$F_{LV}(t) = \int_{t_0}^{\infty} \left[\exp\left(-\frac{(t-s)^2}{\sigma^2}\right) \right] \cdot \left[\frac{\exp\left(\frac{t_0-s}{t_L}\right) - \exp\left(\frac{t_0-s}{t_R}\right)}{1/t_R - 1/t_L} \right] ds$$

$$REF = 1 - \exp\left(-\frac{1}{Pt_R}\right)$$

$$LEF = 1 - \exp\left(-\frac{1}{Pt_L}\right)$$

$$REF = \frac{\text{Ejection Volume}}{\text{Right Heart Volume}} \quad (\text{Right Heart Ejection Fraction})$$

$$LEF = \frac{\text{Ejection Volume}}{\text{Left Heart Volume}} \quad (\text{Left Heart Ejection Fraction})$$

P = Pulse Rate

t_R = Right Mean Transit Time

t_L = Left Mean Transit Time

The utility of this form is found in the following two observations:

- 1) The time interval from half-height to centroid of the RV curve is nearly equal to t_R
- 2) The time interval from half-height to centroid of the LV curve is nearly equal to $t_R + t_L$

Thus the left heart ejection fraction may be determined by two very simple measurements on the RV and LV curves. In Fig. 8 we have plotted a measured LV curve from a post-op heart study (aneurysectomy) and the values predicted by our model using parameters as determined in 1) and 2) above. The width of the gaussian lung distribution could be varied to obtain a best fit but we have found that the value $\sigma = T/4$, where T is the mean central transit time, provides a good estimate. The agreement between the calculated curve and the actual heart curve is surprisingly good, a fact which we feel vindicates the extremely simple way we chose to determine t_R and t_L and hence REF and LEF.

Figure 8 The black dots are data from the LV curve of a cardiac study. The open triangles are the calculated function F_{LV} using the values of t_R and t_L as determined by the simple method described in the text. Systemic flow and recirculation are in evidence for times beyond 20 seconds.

Conclusion

A computer program may be thought of as a meticulously detailed description of a task to be performed by the computer. We have recognized a lack of clear understanding of tasks or an inability to precisely define goals as a source of problems in fields aside from medicine. A common misconception is that the computer is intrinsically useful. Since it is only an inanimate automaton, the user must be able to recognize the tasks to which it is applicable and incorporate the machine into a viable system. We have therefore focused our attention on the definition of clinical problems and the methods by which digital computers may best be used to expand our capabilities in the practice of nuclear medicine.

To this end we have devoted much of our effort toward the development and pursuit of non-invasive techniques for cardiovascular assessment. We are using our scintillation camera/computer system to perform dynamic nuclear angiocardigraphic studies in patients undergoing surgery for the correction of cardiac lesions. Preliminary observations of persistent regional nuclide concentration, abnormalities of contour, abnormalities of flow pattern, and variations in observed mean transit times suggest that such studies may be useful in the evaluation of congenital and acquired heart disease. In addition, work with simple mathematical models has demonstrated the feasibility of extracting such parameters as ejection fraction from activity histograms obtained over the cardiac chambers. We are encouraged that these computer-assisted techniques will provide a new source of clinically useful data, including myocardial function and cardiopulmonary blood volumes in critically ill, or recently traumatized patients, as well as for screening and diagnostic purposes.

References

1. Watson, D. D. and Nelson, J. P., "A Computer for Nuclear Medicine at Mt. Sinai Hospital," Proceedings of Symposium on Sharing of Computer Programs and Technology in Nuclear Medicine. Oak Ridge, Tenn., USAEC Conf - 710425, 147-153 (1971).
2. Oldendorf, W. H. and Kitano, M., and Shimizu, S., "Evaluation of a Single Technique for Abrupt Intravenous Injection of Radioisotope," J. Nucl. Med. 6:205 (1965).

INVESTIGATION OF CARDIAC DYNAMICS USING A
SCINTILLATION CAMERA AND DEDICATED COMPUTER

Richard A. Ponto, Marvin E. Goldberg, Merle K. Loken

Division of Nuclear Medicine
University of Minnesota Hospitals
Minneapolis, Minnesota

ABSTRACT

Investigation of cardiac dynamics has been carried out by using a scintillation camera and a dedicated computer (N.D. Med II). Tc-99m tagged albumin or pertechnetate were used as the radiotracer to investigate cardiac flow patterns, cardiac output, intracardiac transit times, systemic circulation time, and for the detection of cardiac shunting. The dedicated computer was used for recording and analysis of the camera data. Examples of the studies performed and of the type of information obtained is presented.

Evaluation of cardiac dynamics by a simple, non-invasive technique is relatively easily accomplished by using radioactive tracers and the scintillation camera. Over the last few years we have used the dynamic radionuclide study to investigate a variety of cardiovascular parameters. Several modes of recording and analyzing the scintillation camera data have been used. These have included the qualitative evaluation of timed, sequential scintiphotograms and cine-scintiphotography. Quantification of the studies has been achieved by recording of digitized camera data in a two dimensional 1600 word multichannel analyzer with subsequent storage in computer - compatible format on magnetic tape. The computer analyses of data were then handled off line by a CDC -3300 computer (1,2). A Nuclear Chicago data-store playback accessory has also been utilized (3).

Recently we have connected a small, dedicated computer to a scintillation camera. It is our opinion that this type of system will provide the greatest ease and flexibility for use in cardiovascular nuclear medicine procedures. This report will consider our use of the scintillation camera-dedicated computer system for cardiac studies.

Some of the cardiac parameters which can be evaluated by this method include:

1. a flow index from the width of radiotracer dilution curves,
2. cardiac output,
3. systemic circulation time,
4. intracardiac circulation times, and
5. detection of shunting.

The instrumentation used for these studies consists of a Nuclear Data Radicamera interfaced to the Nuclear Data Med II computer. The scintillation camera has a 10 1/2 inch diameter useful field of view. It is fitted with a low energy, parallel-hole collimator or, for studies on infants, a pinhole collimator. A dual channel pulse height analyzer in the camera permits the use of two different radionuclides simultaneously.

The computer system is comprised of a camera interface, 100 mHz dual ADC, computer controller and interface, 12 bit x 16K memory computer with a 2 μ second cycle time, 12 million bit disc for mass storage of data and program storage, magnetic tape transport, CRT terminal, teletype terminal, and CRT display scope (Figure 1). Most of the software used by us for the system is written in a variant of Fortran language (Nutran).

Flow patterns were established by utilizing the rapid intravenous injection of 10-15 mCi of ^{99m}Tc -albumin or pertechnetate. When possible the examination was performed with the patient seated in the left anterior oblique position with respect to the detector head. The detector head was angled caudad at approximately 20 degrees. Recording was carried out sufficiently long to include recirculation or in some cases, to achieve equilibrium of the tracer in the circulating blood.

The dynamic cardiac studies are recorded by calling in a data acquisition program from disc (Figure 2). Input relating to the matrix size, frame rate and total number of records are supplied to the computer via the CRT terminal. When the study is started, the camera's X and Y positional signals and a Z strobe signal are sent to the interface. The X and Y signals are digitized for storage of the event in one of two buffers each with 4096 twelve bit words. The contents of one memory buffer represents one record. At the end of each preselected time interval

Figure 1 Block diagram of camera-computer system

Figure 2 Flow diagram for cardiac studies

the data in memory are transferred to disc and the other 4096 buffer begins to store data. With this flip-flop arrangement there is negligible data loss between records. The transfer from memory to disc requires a maximum of 160 milliseconds. Consequently, the minimum time duration for a frame in the 64 x 64 matrix format is 0.16 seconds. In the 32 x 32 matrix format, each frames duration can be decreased by a factor of four. The disc has a capacity for storing 180 records of data. Magnetic tape is available for long term storage of data.

At the completion of the study, selected frames are recalled from the disc and displayed on the CRT. Regions of interest are then assigned by using a light pen, X-Y coordinate markers, or with a display Z control containing upper and lower level discriminators. A maximum of 12 regions of interest can be assigned with this system. With the cardiac studies, one of the regions of interest includes the entire field and its data are used to correct for the dead time of the camera-computer system. After selection of the regions of interest, all records are sequentially recalled from disc so that counts per frame in each region of interest can be determined. The results are then stored on disc. Programs can be called to smooth the curves, correct for dead time, and display curves on the CRT. In addition, the curves can be plotted on the teletype along with appropriate numerical data. Selected data points from the curves can be recalled from disc and used in the determination of various cardiac parameters.

A parameter that we have found to be particularly useful is the change in width of the dilution curve (full width half maximum-FWHM) as the radioactive bolus passes through the cardiac chambers (Figure 3). All values are normalized in width to the right ventricle to correct for variations in the manner in which the bolus is injected. This parameter is a sensitive indicator of the rate of transport of the tracer through the cardiovascular system. For example, normal subjects exhibit FWHM $\frac{\text{Lt-atrium}}{\text{Lt-ventricle}}$ and FWHM $\frac{\text{Rt-ventricle}}{\text{Lt-ventricle}}$ ratios of less than 2.0,

whereas $\frac{\text{Rt-ventricle}}{\text{Lt-ventricle}}$ patients with valvular disease have values greater than 2.5 (3). For this analysis a simple algorithm is used which finds the peak value of the curve and data sets representing the one-half peak values. The time interval between the two 50% values is then determined. Occasionally it is necessary to extrapolate the curve to the base line in order to make these measurements. It is often helpful if the data are smoothed before determining the FWHM. This can be accomplished by passing a smoother through the curve or by condensing frames. The latter approach is useful when the frame rate is greater than two per second.

Cardiac output determinations can also be made from the dilution curves. Generally the right ventricle curve is utilized for this purpose. The area under this is easily determined by summing the counts from each frame up to the point of recirculation (Figure 4). The downslope of the exponential curve is extrapolated to the base line in a conventional manner and the area (B) under the remainder of the curve is estimated by $B = C_b / \lambda$ where C_b represents counts in the data set at the start of recirculation and λ is the slope of the exponential curve (determined by $\lambda = .693 / T_{1/2}$). The count rate at equilibrium, C_f is established and the blood volume determined. The cardiac output is then calculated according to the formula shown in Figure 4. A frame rate of 2 per second was found to be adequate for most cardiac output studies.

Figure 3 Determination of full width half maximum value for dilution curve.

Figure 4   Determination of cardiac output

Systemic circulation time is also a useful index of cardiac dynamics. Systemic circulation times are determined by measuring the time interval from the peak of the right ventricle dilution curve to the peak of the recirculation on the same curve (Figure 5). Intracardiac circulation times such as the transit time from right to left ventricle can be determined in a similar manner and likewise provide useful information on cardiac function (1).

The detection of intracardiac shunts is accomplished by analyzing the dilution curves representing the passage of radioactivity through selected regions of the lungs (Figure 6). It has been shown that the downslope of the lung curves will be prolonged if there is significant shunting (4,5,6). Using the technique described by Folse and Braunwold (4) and adapted to the scintillation camera by Rosenthal (5), the ratio C_2/C_1 is used as an indication of the slope of the clearance curve. The time to reach the peak count rate, C_1 , is T_1 and C_2 is the count rate at $2T_1$. An example of a curve from a patient with a left-to-right intracardiac shunt is shown in Figure 7. The C_2/C_1 ratio in this case is 0.60 in contrast to the value obtained from the normal subject whose C_2/C_1 ratio was 0.38 (Figure 6). We are currently studying a series of patients to determine normal values for this parameter.

For heart studies, the small, dedicated computer is used with ease and flexibility with a scintillation camera. A number of examples showing the type of data which can be generated has been illustrated. There are a number of other parameters which can be examined by using this instrumentation and for which we are currently developing techniques to study. These parameters include 1) ventricular volumes, 2) stroke volume, 3) ejection fraction, 4) myocardial blood flow and 5) quantification of shunts.

An important feature of a dedicated computer for use in Nuclear Medicine is the capability of easily interacting with the system during data analysis. This is important in helping with optimal assignment of regions of interest and in determining the best routes for data analysis.

The availability of a low cost disc adds considerably to the efficient use of the system described in this report. The disc is used to store moderately large quantities of camera data and it permits the use of relatively high frame rates. Software is also stored on the disc and programs can be accessed within seconds when needed for a particular operation. Files of processed intermediate data, such as counts per frame in the region of interest, are also stored on disc. These files can be rapidly accessed as needed for various analyses. The magnetic tape provides the means for long term storage of studies and for recording of files containing large quantities of information, i.e., more than the capacity of the disc (180 records). In addition, the tape unit can serve as a link in communications with other computer systems. For example, we are presently considering the preprocessing data on the Med II computer and writing it on magnetic tape. The tape can then be taken to the CDC-3300 computer at the Biomedical Computer Center for certain complex analyses which are not feasible on the small computer. The use of special plotters and high speed printers at the large computer center is also being considered. Again the tape can serve as the link between the dedicated computer and the large computer center.

The large memory capacity of the Med II computer allows 8 K of core to be used as buffer storage for incoming camera data. By dividing this into two 4 K buffers, a flip-flop data storage

Figure 5

Determination of circulation

Figure 6

Determination of C_2/C_1 ratio from lung curve of a normal subject.

RIGHT VENTRICLE

RIGHT LUNG

Figure 7 Right ventricle curve and lung curve in a patient with left-to-right shunting.

system is utilized and rapid frame rates with negligible data loss can be achieved. The remaining 8 K of core provides considerable storage for software thus increasing the flexibility of the system. At the present time most of the software needed for the basic operation of this system is provided by the vendor. The capability of programming in a high level language allows the user to easily add the necessary programs for specific operations as needed. The user can utilize assembly language programming as the need arises and as his expertise for this type of work is developed.

In summary, a small dedicated computer-scintillation camera system for use in cardiovascular nuclear medicine procedures has been described. Examples of the types of information obtained has been shown. It is expected that this type of system will prove to be useful in all dynamic radionuclide studies as well as in static imaging.

REFERENCES

1. M.K.Loken, R.A. Ponto, R. Bache and H. Burchell, "Intra-venous Radioisotope Angiography with Computer Processing of Data " Am. J. Roent., Rad. Ther. and Nuclear Medicine 112, 682-690, (August, 1971).
2. R.A. Ponto, G.S. Kush and M.K.Loken, "The Use of a CDC-3300 Computer for Processing Scintillation Camera Data in Nuclear Medicine" In Proceedings of Symposium on Sharing of Computer Programs and Technology in Nuclear Medicine, Oak Ridge, Tennessee, (April 2-3, 1971).
3. M.E.Goldberg, R.A. Ponto, and M.K.Loken, "Radioisotope Angiography" to be published in Radiology.
4. R. Folse, E. Braunwald, "Pulmonary Vascular Dilution Curves Recorded by External Detection in the Diagnosis of Left to Right Shunts ", Brit. Heart J. 24, 166-177 (March, 1962).
5. L. Rosenthal, "Nucleographic Screening of Patients for Left to Right Cardiac Shunting" Radiology 99, 601-604 (June, 1971).
6. N.P. Alazraki, W.L. Ashburn, A. Hagen & W.Friedman, "Detection of Left to Right Shunts with the Scintillation Camera Pulmonary Dilution Curve", J. Nuc. Med. 13, 142-147 (Feb, 1972).

INCREASING THE INFORMATION CONTENT IN A CRT DISPLAY
FOR A NUCLEAR MEDICINE COMPUTER SYSTEM

Philip A. Miller
Donald R. Strange
Arthur F. Shufelt

ABSTRACT

An on-line display system is described which can display a 64 x 64 data matrix in a 128 x 128 format. Either four 64 x 64 images may be displayed or a single 128 x 128 image with the additional points interpolated from the raw data. The four images may be separate views of the same study or four views of the same image with different contrast, background suppression, etc. Alpha-numerics may be superimposed on the display to identify patients, counts, or provide comments. The hardware configuration requires less than 25% of the processor time to display a 60 hertz/second flicker free image. Data may be manipulated while it is being acquired.

Medical Data Systems Corporation
A Warner-Lambert Subsidiary
2300 Fisher Building
Detroit, Michigan 48202

The basic function of the display is, of course, to present digital information visually. In some ways this is the most important function of a nuclear medicine system since it is a primary, and sometimes the only output.

Our goal has been to develop methods of display which take full advantage of the versatility of computer technology, while keeping the function of the system simple and straightforward. A physician or technologist must be able to use the system with a minimum of computer training. The best utilization of a nuclear medicine system is for the physician to deal with it directly, glean the maximum information possible from the display.

We feel that a computer can do more than merely display images, but can maximize the information available visually. A computer can generate many more representations from a single scan than it is practical to photograph, more than one of which may be clinically valuable.

In an image of an organ, the crucial factor is to be able to differentiate an abnormality. By using a flexible, software controlled display, full intensity differentiation can be used to enhance a subset of the total count range. For example, a brain scan may have a range of counts from 0 - 7000 counts/channel. The area of clinical interest in the brain, however, may have a range from 100 - 300 counts/channel. If this count range is set to correspond to full intensity modulation, much greater differentiation is achieved.

Although we are acquiring data in the standard 64×64 matrix, we feel that a 64×64 display is not the most helpful. It is difficult to see detail in an image because of the graininess of the display. Ideally, to see detail one would enlarge the image; but while restricted to a 64×64 display this merely increases the graininess.

We have employed a 128×128 display, which we feel overcomes some of these problems. By expanding the display matrix the image looks more coherent to the eye. This, of course, does not increase resolution; but does give a more pleasing representation of an organ.

The 128×128 display can also be used to give a larger image. With the same dot density, or graininess, of a 64×64 image on an 8×10 cm. scope, the expanded matrix can be $6'' \times 6''$. A life size display can also be utilized, showing the display matrix as a $10'' \times 10''$ square.

As a result of using a 128×128 display, four 64×64 images can be displayed simultaneously. This can be used to show one image with four degrees of contrast enhancement, four views of one organ, four frames of a dynamic study or any other combination.

We have also developed character generation; displaying alphabetic and numeric characters on the CRT.

Fig. 1. Block diagram of display hardware.

Figure 1 is a block diagram of the hardware used for the display. An image displayed on the CRT's is generated in the computer in the form of digital data. This data is translated into an image by a display controller connected to the computer by high speed data channel. This means that the display controller, once given certain parameters from the computer automatically scans the screen, generating an image of the data. This process is independent of the computer and utilizes only a small part of the computer's real time capability.

The display controller generates a raster scan with a variable size matrix. Each point of the matrix is intensified to one of 16 levels. The X, Y, and Z outputs can be connected to multiple display scopes.

We are currently using a 12" x 14" viewing scope, and an 8 x 10 cm. photographic scope.

Fig. 2. Top image: 64 x 64 matrix display.
Bottom image: Same image, 128 x 128 matrix display.

Fig. 3. 128 x 128 matrix display of lung.

Fig. 4. One lung scan displayed simultaneously with four levels of contrast enhancement.

Fig. 5. Four simultaneous, different views of a brain.

The 128×128 display is used to display 64×64 data by interpolating three additional points for each channel of ADC data. The interpolated points are the average of nearby real points.

Fig. 6. 128×128 display of a liver with title and total count accumulation superimposed.

Fig. 7. Example of alphanumeric display.

Alpha-numeric characters can be generated on a CRT by using a 5 x 7 dot matrix for each character. All upper case letters, numbers and punctuation marks can be represented in this manner. With this capability the CRT display can be used in place of teletype printout for communication. This gives the advantages of being instantaneous, silent, and utilizing the same CRT which is used for images. Thus textual identification can be superimposed on an image, or the entire screen can be used for text. This may prove to be very helpful when used with a light pen for operator interaction.

In conclusion, I have described a method of computerized image display which offers complete software control of the display parameters. The 128 x 128 matrix provides a more pleasing representation of a single image, and gives the capability to display multiple views. Alpha-numeric titles can provide self-identifying photographs.

The software approach will allow display parameters to be pre-programmed for specific types of studies, approximating the optimum contrast while giving the user ultimate control.

MATCHING GRAY SCALE DISPLAY PROGRAMS TO VARIOUS FILM TYPES

James C. Carlson, M.S.

Radiology Muskegon, P.C.
and
Nuclear Medicine Department
Hackley Hospital
Muskegon, Michigan

ABSTRACT: Optimum computer produced photographs are obtained when the display program produces oscilloscope display luminance ranges appropriate to the characteristics of the film employed. This optimum luminance range was determined for 6 film types (Kodak 5069, S0-392, blue brand, and plus X; Dupont SF2; Polaroid 107).

Professional photographers can create exciting detailed prints where an amateur produces dull hazy prints because the professional understands the relationship of subject lighting to the characteristics of his film. When creating computer generated photographic images it is equally important to understand the relationship of subject lighting to film characteristics. The results of an effort to optimize the relationship of the oscilloscope display program to several types of film will be discussed after an introduction to some of the basic principles of photography.

In normal situations photographic contrast of the subject is determined by subject material and lighting conditions. To produce a high quality picture a film must be chosen to match these factors. In computer programming of oscilloscope displays the programmer has complete control of the subject conditions. He may either choose a film to match his display program or he may write his program to match the characteristics of the available film. In each case improper knowledge of the film characteristics can lead to the use of a completely inappropriate display program. (2) This is illustrated in figure 1.

Figure 1. The luminance range of the display must match the characteristics of the film. A. This view of Polaroid 107 film was obtained from a computer display program appropriate to the film characteristics. B. On the same film a display program with a luminance range excessive for the film produces an image of inferior quality.

FILM CHARACTERISTICS

Attention is directed to the essential film characteristics of density, latitude and contrast. (1,3) Spectral sensitivity and resolution are not considered. Film evaluation is determined by the transmission characteristics of density variations on the particular film base. It should be noted that the principles discussed also apply to the reflected light of polaroid pictures.

Film density is the logarithm of the ratio of incident to transmitted light expressed as:

$$\text{Film density} = \log \frac{\text{incident luminance}}{\text{transmitted luminance}}$$

A density of 0.3 corresponds to a film area where the transmitted luminance is one half of the incident luminance. Likewise an 0.3 change in density represents a 50% change in luminance; e.g. the luminance where the film density reads 1.0 is one half that where the density is 0.7. Illustrating further, a density change of 1.0 represents a 10X change in luminance and a density change of 2.0 represents a 100X change in luminance.

A typical sensiometric (H & D) curve and some of its salient features are illustrated in figure 2.* Latitude is generally considered as the total exposure range which produces useful densities on a negative. The definition of latitude shown in figure 2b, however, is the amount by which the exposure can deviate from an ideal exposure while still obtaining a satisfactory negative.(1) This distinction in definitions will be important later when latitude is used to describe the relationship between the computer program and film type and the latter definition is used.

*Note: In each of the curves shown in figures 2 through 4 all or some of the following information is missing; film type, exposure light frequencies, development time, developing chemicals, and development temperature. Without this information none of the curves demonstrated may be considered performance characteristics guaranteed by the manufacturer.

Figure 2. The salient features of sensimetric curves are illustrated. Useful film densities are obtained when the luminance of the subject ranges from that corresponding to the toe to that of the shoulder of the curve. The film *gamma* is the slope of the straight portion of the curve; whereas, contrast index is the slope of the line joining the minimum and maximum useful densities. Latitude, as shown here, is the unused portion of the total useful exposure range.

Figure 3. Typical sensimetric curves for 4 different films. Plus X is a film normally used in conventional photography. Its low gamma is suitable to general photography where the luminance range of the subject extends over 2 decades. The other films are all high contrast films, as evidenced by their large gamma values. Type 5069 requires very high exposures, type S0-392 is more sensitive and type 1521 is the most sensitive and most granular.

Figure 4. These typical curves illustrate the great sensitivity of this film to development time. The gamma of the film changes from 2.2 for a development time of 45 seconds to 6.4 for a development time of 8 minutes. Notice that a given contrast ratio of 3.8 in the subject will produce a density change of 0.90 for a 1 minute development and a density change of 1.3 for a 3 minute development.

Additional sensiometric curves provided by film manufacturers are shown in figure 3. These curves have different gamma values, different exposure values at the toe of the curve, and different maximum densities. These parameters are modified by many conditions, the most significant being development time and chemicals. An example of the dependency of the sensiometric curve on development time is demonstrated in figure 4. (1)

Contrast and contrast ratios are terms describing differences in luminances. These terms are defined by the following formulas.

$$\text{contrast} \quad C = \frac{B2-B1}{B2}$$

$$\text{contrast ratio} \quad C_R = \frac{B2}{B1}$$

where: B1 - luminance of darker object
B2 - luminance of brighter object

FILM TESTS

Determination of the optimum display contrast ratio began by using manufacturer's film sensiometric curves as guides to the probable display contrast ratios required. The actual desired contrast ratio was then determined experimentally in conjunction with the display program, oscilloscope intensity setting, film types, and development procedures.

For this purpose a program was written to produce on a Tektronix RM503 oscilloscope 12 squares, each of which represented a step on a gray wedge. The luminance levels required were obtained by altering, according to a table, the number of times each dot was displayed. Several tables were constructed with various values for the total luminance range or total contrast ratio. Tables

with "total contrast ratios" (contrast ratios determined from the maximum and minimum steps) of 8, 10, 13, 16, 21, 35, 83, 126, and 166 were constructed. One of these tables appears in figure 5 together with some analytical data concerning the display numbers.

Square	Decimal number	Fractional change	Change in log of No.	Negative octal
1	10	--		7766
2	13	1.30	.114	7763
3	17	1.31	.106	7757
4	22	1.29	.112	7752
5	28	1.27	.105	7744
6	36	1.27	.109	7734
7	46	1.28	.107	7722
8	59	1.28	.108	7707
9	76	1.29	.110	7664
10	98	1.29	.110	7636
11	126	1.29	.109	7602
12	162	1.29	.110	7636

Contrast ratio $\frac{162}{10} = 16$

Figure 5. This is an example of the development of a display table. An attempt is made to select decimal steps that produce equal changes in the log value; i.e. the luminance change per step should be a constant percentage. The computer display table requires the negative octal equivalence of the decimal display number.

Test patterns were produced on 6 different films using several tables. Not all tables were applied to all films but each film was exposed to several tables which, according to sensimetric curves, could be expected to produce a film extending over the density range of 0.05 to 2.0. In each case the oscilloscope intensity setting was varied to produce a film with the lowest gray step just within the range of minimal perceptible contrast. The films were reviewed to select those where the contrast of all steps was perceptible.

An attempt to demonstrate this selection process is shown in figure 6. (The word attempt is used to emphasize that the original material may not be faithfully reproduced in this manual.) In this series of test patterns made on Polaroid type 107 film the "total contrast ratio" is inadequate (figure 6a), optimal (figure 6c), and excessive (figure 6d).

The largest "total contrast ratio" producing a film satisfying the above criterion is listed in figure 7. Since this selection process requires subjective evaluation, the contrast ratio given in figure 7 could be very observer dependent and, according to basic principles, is dependent on the development process. Film images produced using the appropriate contrast ratio should all appear identical to the viewer.

Figure 6. Test patterns obtained using displays with "total contrast ratio" of 13, 21, 35, and 83. Figure C of the square patterns was selected as the optimum image. The optimum image of the series of patient pictures is more difficult to select.

FILM TYPE		OPTIMUM CONTRAST RATIO
KODAK	HIGH CONTRAST COPY 5069	13
KODAK	BLUE BRAND (X-RAY)	35
KODAK	PLUS X	166
KODAK	SOLAR FLARE S0-392	13
DUPONT	SF2	35
POLAROID	107	35

Figure 7. The oscilloscope display program should be written with a "total contrast ratio" or luminance range appropriate to the film being used. The appropriate range determined in these experiments is listed above. The development conditions for these films were: 73°F; 5069, D-19, 4 min.; Blue Brand, Xomat processor, 90 seconds; plus X, D-76, 7 minutes; S0-392, D-19, 2.5 minutes; Dupont SF2, D-19, 3 minutes.

COMMENT

An unstated assumption is that the most desirable film for clinical viewing is one where the gray scale ranges from minimum detectable to maximum readable under normal viewbox lighting; i.e. the best film is one with very little or no latitude. The validity of this assumption seems self evident considering that there is a minimum contrast perceptible to the eye and that certain areas of films of low density range may not exceed this minimum contrast; whereas they may exceed it when the film has a very long range of densities.

The assumption that the best film has no latitude is repugnant to the professional photographer because normally he adjusts his picture density range by adjusting his paper development procedures. While it could similarly be said that there is no need to produce computer images with no latitude since paper picture records could be made to provide the desired latitude, this method seems unduly burdensome when correct computer programming will suffice. By extension it may be said that the reputation of Polaroid film for poor gray tones is erroneously described as a fault of the film rather than a fault of the display technique.

For purposes of this presentation it has been assumed that the luminance is directly proportional to the number of times the dot is redisplayed and no tests have been made to confirm this assumption.

At the present time Kodak high contrast copy film #5069 is used for all 35mm pictures at Hackley Hospital because it is very convenient to develop by hand using small developer tanks. (Many factors preclude our use of an Xomat automatic processor). A disadvantage of this film is its slow speed. Since SO-392 has a similar contrast index and is a faster film, its possible replacement of 5069 is being considered.

SUMMARY

Pictures produced from the computer oscilloscope should have no latitude in order to improve the probability of detecting minimal density changes. In order to produce images of this type either the program must be matched to the film characteristics or the film must match the program. Six film types were tested to determine the optimum "total contrast ratio" to be used in the computer display program. When the computer program is designed to match the film, the choice of film to use may be based on other factors.

REFERENCES

1. Kodak Professional Black and White Films, First Ed.; Eastman Kodak Co., Rochester, N.Y. 1971
2. J.C. Carlson, "Problems of the Clinical Nuclear Medicine Facility in the Utilization of a Dedicated Minicomputer", Proceedings of Symposium on Sharing of Computer Programs and Technology in Nuclear Medicine, CONF-710425, Oak Ridge, Tennessee, April 2-3, 1971.
3. H.R. Luxenberg and R.L. Kuehns, editor, Display Systems Engineering, McGraw Hill Book Co., New York, N.Y. 1968

COMPUTER CONTROLLED, INTENSITY MODULATED CRT DISPLAY
OF ISOTOPE SCAN DATA

Robert O. Smith

ABSTRACT

A Digital-to-Analog (D/A) converter design is described which has the capability of intensity modulating each location of a cathode-ray tube (CRT) X-Y display. For such a display to function under computer control, proper summing of the D/A converter output and computer CRT unblanking pulses is required. This has been accomplished and is a part of the complete design.

Department of Medicine
University of Mississippi Medical Center
2500 North State Street
Jackson, Mississippi 39216

COMPUTER CONTROLLED, INTENSITY MODULATED CRT DISPLAY OF ISOTOPE SCAN DATA

Robert O. Smith

Department of Medicine, University of Mississippi
Medical Center, Jackson, Mississippi 39216

For displaying isotope scan data on a cathode-ray tube (CRT), even under computer control, variation of the number of dots seen has been the primary mode of indicating the various isotope concentrations present in the scanned area.¹⁻² In an effort to obtain photographable displays, CRT beam defocusing has been used but does not produce acceptable results. The more dots that the computer is required to display, the slower the display repetition rate must be. For this reason, the dot density technique is of value when one can only view the CRT screen. However, for photographic purposes, a better solution seemed possible by varying the intensity or brightness of all the dots that could be produced by a particular CRT display system. An intensity modulated raster display would be seen of the type so familiar to television viewers. Such computer controlled, disc storage and display systems are available commercially. Cost alone can prevent use of such a system with a small laboratory type of digital computer (mini-computer).

A smaller scale simulation has been designed, tested and is in use. Though digital to analog (D/A) converters with sufficient speed are available commercially, a special output stage was required for intensity modulation because a CRT unblanking pulse is synchronously provided by the CRT deflection circuits of most computer driven displays. In the present instance, a positive 12 volts will completely blank the beam regardless of the intensity control setting. A negative 12 volts will turn the beam on with maximum brightness being limited by the intensity control. For proper modulation, the intensity axis of the X-Y display must be at a positive 12 volts to cut off the beam whenever it is to be moved or the computer is calculating. For the 4 microseconds that the beam would be on to produce a dot, the voltage should be constant between zero and some negative value less than 12. This negative voltage must be directly related to the numerical value to be displayed. A capacitively coupled intensity input for a X-Y display is satisfactory where intensity modulation is not required but is completely inadequate for intensity modulation. Capacitive coupling of the unblanking pulses causes a d-c bias shift in the CRT. In order to meet all the requirements, it was decided to design the complete circuit.

Figure 1 is the circuit of the D/A converter with its complementary follower output. The computer binary bit signals come in through connector J901 on the left. This is a 5-bit or 32 level converter so the input buffer

COMPUTER OUT
HP 12566 - 6002

ABOVE SECTION DUPLICATED FOR
EACH BINARY BIT.

All diodes are IN 645

Figure 1. Connector J901 (on left) connects the computer output to the converter input. The weighted resistor network (1K, 2K, 4K, 8K, and 16K) is seen at the center. A Hewlett-Packard Model 12555A Digital-to-Analog Converter supplies the X-Y display deflection and unblanking signals. The unblanking signal synchronizes the intensity modulation output with the computer. The complementary emitter follower output uses a 2N3735 and a 2N3763. These high-frequency and medium current transistors are used to reduce cable capacitance problems.

amplifier and switch section is duplicated four additional times. The computer in use is a Hewlett-Packard 2115A and the 12566-6002 Input/Output printed circuit board is one identified as a microcircuit interface board. Transistor-transistor logic (TTL) is used on the interface board and the two diodes in the base circuit of the 2N3638 insure that this point shall never go very negative. Minor changes in the input buffer amplifier will allow other input levels to be used. In this amplifier, when the most significant bit (MSB) is one, the positive input voltage will be inverted to negative by the buffer and will turn on the 2N2905 transistor switch and apply the negative 10 reference voltage to the MSB ladder resistor. At the same instant, the 2N3053 transistor will be turned off. If the MSB is zero, the reverse condition exists in that the 2N2905 is off and the 2N3053 is on. This is conventional for such a weighed-resistor configuration.³ This three transistor circuit is duplicated for each of the binary bits from the computer.

The zero to negative 10 volt output of the converter is applied to the base of the second 2N2905 transistor through the one thousand ohm resistor. For other transistor/diode combinations, a change would likely be required in the resistance of the 500 ohm, 5 watt resistor. With this combination, when the computer output is zero and the unblanking is a negative 12 volts, the complementary emitter follower output is also zero. If the computer output is a decimal 31, the complementary emitter follower output is a negative 10 volts. For the numbers between zero and 31, a relative negative level is available. However, when the unblanking is a positive 12 volts, the follower output also will always be approximately 12 volts and the CRT beam will be blanked.

Figure 2 shows some additional circuitry which was added to interface to the X-Y CRT displays which are in use. The Hewlett-Packard Model 1330A has a P4 (white) phosphor for photographing through color filters. The X, Y and Z inputs are differential and can be connected for proper signal polarity. A 10 K ohms potentiometer controls signal amplitude to the intensity input of each X-Y display.

The Hewlett-Packard Model 1331C is a variable persistence or storage display and is used for computer communication with the operator. Though its intensity can be modulated, only about 4 levels can be seen in the storage mode. Successive single sweeps can be used to show build-up of areas in a scan. Selected by a switch, its intensity input can be either from the D/A converter or an adjustable intensity control. This last control was added so adjustments could be made from the front of the console instead of going to the back whenever an adjustment was desired. The Model 1331C has an inhibit input which can be used to inhibit writing or erasing. This allows the same deflection signals from the computer to drive both displays. When the computer has finished communicating with the operator, the storage display can be inhibited and the last message can be retained.

Though black, white and ten shades of gray have been considered to be about maximum resolution for the human eye, 32 levels were chosen for this converter because the additional levels might provide a more pleasing display. Figure 3 is a Polaroid photograph showing levels zero and 4 through 31. This pattern was chosen so that the smooth level transitions might be seen. This is a 256 x 256 dot display in which 8 x 8 dots form each element in a 32 x 32 array.

The computer program which drives these devices is called "RANDD" which is an abbreviation of "read and display" paper tape. The program will

Yellow test

J 902

Output of
complimentary
emitter follower

Figure 2. Addition circuitry used to provide drive capability to two X-Y displays. A Hewlett-Packard Model 12555A Digital-to-Analog converter provides the deflection and high unblanking signals. Test jacks are provided for inspection of the various signals.

read in and display a 32 x 32 array with options. The first option is whether to scale the data from zero to 31. If the maximum number in any element is more than 31, then obviously, scaling is required for a correct display. Scaling is generally used because the intensity setting can then remain the same for all displays.

Transposition of the data to any one of four different orientations is the second option. This is available in order that the photograph can be standardized for any type of scan orientation. For photographing multiple images through color filters, the slow speed of the color film has required that 64 sweeps be done for each exposure. The program handles this automatically if switch register bit number 15 is set to zero. If this switch is set to one, a single sweep is made each time the computer run button is pushed. One sweep is satisfactory for black and white photography.

Using the equipment described, the picture in Figure 4 is from a normal human myocardial scan. Figure 5 is an abnormal scan of the human myocardium.

ACKNOWLEDGEMENT

Supported in part by Grant HE 07628 from the National Heart and Lung Institute, National Institutes of Health.

I am indebted to Mrs. Margaret Boyd for writing the computer program.

REFERENCES

1. Jim Baker, "Display Programs for the PDP-12", Proceedings of Computer Processing of Dynamic Images from an Anger Scintillation Camera, Vol. 1 Chap. 15 (Washington Univ., St. Louis, Jan., 1971).
2. Franklin Borkat, "Visual Display for Digital Computer Information", J. Nucl. Med., 16(6), 341 (June, 1971).
3. D.F. Hoeschele, Jr., Analog-To-Digital/Digital-To-Analog Conversion Techniques, pp. 116-119 (John Wiley & Sons, Inc., New York, London and Sidney, 1968).

Figure 3. A Polaroid photograph of 32 x 32 array using intensity modulation. The levels shown are zero and 4-31. Levels 1, 2 and 3 were intentionally omitted to provide the staggered line display. The linearity and resolution are made more obvious.

Figure 4. Computer controlled intensity modulated display of human myocardium considered to be normal. The isotope was potassium-43.

Figure 5. Using the technique and isotope of Figure 4, this is anterior view of an abnormal human myocardium.

SHARING OF ALGORITHMS FOR SCAN PROCESSING IN NUCLEAR MEDICINE

A. E. Todd-Pokropek

ABSTRACT

It is suggested that the most suitable computer system for a nuclear medicine department is larger than the small dedicated gamma camera computer system that is found in many centres. For such a system, a survey of the software that has been exchanged, or should be exchanged, is given. A number of suggestions for developing such systems are given, with particular reference to the applications programs (image processing algorithms, data handling algorithms, and display algorithms) to the data base, to the structure of the operating system, and lastly to the possibilities of developing a Nuclear Medicine language.

Institute of Nuclear Medicine
The Middlesex Hospital Medical School
Mortimer Street
London, W.1.
England

and

Institut Gustave-Roussy
16 bis. Avenue P. Vaillant Couturier
94 Villejuif
France

Sharing of algorithms for scan processing in nuclear medicine

To start, it is necessary to examine the environment in which the algorithms are to be shared.

Overview

It could be said that there exists, or has existed, two schools of thought in nuclear medicine scan processing: those who have used large computers to perform complicated manipulations of usually 'static' images, and those who, using a PDP8 or equivalent, have sought to exploit the potentials of their equipment to produce (rather limited) systems primarily devoted to Gamma camera dynamic images. The former group tended to write in high level languages, usually FORTRAN, and the latter were forced to write in low level languages. A certain limited exchange of software has been possible inside these schools, usually between users of particular computers, but virtually none between the two schools themselves.

However at this point in time most of the centres which were formerly using the batch processing computers have obtained small computer systems of their own, while many of the centres using PDP8s, etc., have expanded their systems so that in practice this division is no longer valid. It should surely be possible at this stage to pool our combined knowledge to produce genuinely useful systems. Unfortunately, the field has had a number of enthusiastic amateurs who have succeeded to date rather more in educating themselves in computer techniques than in producing viable systems. It is time for a more professional approach.

What, then, should be the configuration of a typical (potentially) useful system? It seems more reasonable to spend at least £100k on a bigger system, and it is my opinion that a dedicated Gamma camera computer system is not valid. To spend £50k on the equivalent of a sophisticated Polaroid Camera, or data replay system, is a very expensive hobby. For example, the software for a scan processing system is likely to be at least 100k bytes, and therefore a fast backing store is a prerequisite. An archive system is also essential, and therefore some suitable long-term backing store is required. The system must be capable of controlling a large number of data input devices, and therefore a 'real time operating system' is required. In addition, it is likely that on certain occasions it will be more efficient to provide the additional computing power and backing store that the system might require by means of a link to a larger (more central) computer. The system must not be allowed to grow too much, but a further refinement that I anticipate is that the Gamma camera will be isolated

from the N.M. system itself by means of something akin to a small satellite communications computer, or, perhaps alternatively, a rather sophisticated CAMAC-type interface. This should prevent the camera from ever saturating the computer and prevent the system from becoming over directed towards Gamma camera data collection. This system should be capable of servicing the whole nuclear medicine department.

This, in loose terms, has described the hardware. What does the software look like?

Firstly, it must be modular. Secondly, it is bound to be very dependant on the structure of the operating system used. For example, in any such system, all input and output must be performed via the operating system. It will also be necessary to isolate the user of the system from the individual routines themselves by means of some nuclear medicine (higher level) language, or control structure. The modules themselves are likely to be a mixture of high level and low level language but, hopefully, will tend to be written more and more in a pure high level language. Nonetheless, the system will require continuous modification and updating, which should presumably be possible in a transparent manner, that is, without disturbing the system from the user's point of view. Therefore, it should become easier to exchange software, and there should be somebody available to do any necessary modifications.

It is not our task to write the operating system, or at least it should not be. These are bound to be very machine-dependant, and should be provided by the manufacturer. Where cooperation is possible is in the applications programs, in the N.M. language, and most important, in ensuring that the overall structure of the system (modules, operating system, user language, data base) is consistent.

Image Processing Algorithms

One field where cooperation has been possible is in the development of image processing algorithms. However, there is no doubt that many of the existing techniques are of little interest except for use on a machine at least as big as an IBM 360/50, which, as may be seen in the discussion above, is most certainly not the kind of computer that we have been talking about. For example, very few small systems possess hardware floating point arithmetic, and there is little doubt that a (less accurate) calculation in fixed-point arithmetic is of much greater interest to most current users. One of the projects currently being pursued at the Institut Gustave-Roussy, Paris, and at the Middlesex Hospital, London, is the development of a Wiener filter suitable for use on a mini-computer. The first requisite for this is the development of a two-dimensional fast Fourier transform using integer arithmetic only. A further shortcut that is anticipated is that, instead of calculating the filter for each individual case, a library of filters will be generated, and the closest approximation to the true filter selected. This presupposes that the technique is sufficiently stable, which is not certain at this point in time. An illustration of a Wiener filter in the frequency domain is shown in

Fig.1. As a further aid, a series of 5x5 filters (in the space domain) has been generated, using integer values only, which resembles a family of Wiener filters of given 'size' but variable 'amplification'. These are illustrated in the frequency domain in Fig.2. For comparison, Fig.3 shows in the space domain four more filters: the conventional 9-point smoothing, the equally conventional $\{1,2,1 / 2,4,2 / 1,2,1\}$ filter, and two more 5x5 low pass filters. Any user who possesses a convolution 'module' could presumably implement these Wiener-like filters without problems. The development of non-linear, space-variable filters is also under way. One such filter, called the variable shape filter, was developed jointly by the Middlesex Hospital and the University Hospital of Lund (Sweden), and is suitable for use on a mini-computer [1]. The flow chart of the algorithm is illustrated in Fig.4. It has been found to take only twice the time required for a 9-point smoothing on a particular mini-computer. However, it has been found that the low-level language program as such was of little interest even to other users of the same mini-computer as a result of operating system incompatibilities. The FORTRAN version of the program was of much greater interest since the conversion from FORTRAN to assembler was in fact easier than from assembler to assembler.

Another technique of value may be the use of the Hadamard transform, preliminary results of which have been presented in a separate paper. The interest in this transform is primarily in the fact that it is at least ten times as fast as the fast Fourier transform on most computers and is particularly suitable for use with a mini-computer. It is to be hoped that processing techniques, for example, the Wiener filter, will be modified in such a way that the Hadamard transform might be utilisable in place of the Fourier transform. It may be noted that the use of the Fourier transform for the production of the Wiener filters [2] is itself a considerable improvement on the previously published techniques which involve matrix inversion [3], [4].

Image Handling Algorithms

A second field of some interest is that of image handling. By this is meant all the data handling required for both static dynamic image manipulations plus the archiving systems. On these topics there has as yet been very little interchange of algorithms, perhaps because these have been much more the province of the PDP8 school and therefore much more machine-dependant. An attempt was made in the Varian Gamma Camera Users Group to define a structure for an image such that both the images and the algorithms for handling these images, could be readily exchanged. This structure is outlined in Fig.5, but as yet very little interest has been expressed in it. As always, there is a divergence of interest from a general flexible structure to a specific and restrictive system tailor-made to the requirements of a particular centre. This applies equally to data bases and programs, and should be discouraged, particularly as the present generation of computers provides sufficient power to make use of more general systems.

Fig. 1. A contour map and an isometric projection of a true Wiener filter in the frequency domain.

Fig. 2. Four 5x5 filters with increasing negative components giving a family of Wiener-like filters. This is one quadrant in the frequency domain. Note the increasing side lobes.

Fig. 3. Four filters, also in the frequency domain, for comparison with Fig. 2.

Fig. 4. The variable shape filter flow chart. 'Ring' is the data used by the algorithm to determine the step direction for the next count value. (i.e. from one to eight in a clockwise direction). The filter should then spiral outwards.

STRUCTURE OF SCAN FILE

1. Address of SCAN DATA field.
2. Matrix size, across, down.
3. Element size, across, down.
4. Address of PATIENT IDENTIFICATION field:-
Name, number, age, sex
5. Address of COMMENT field.
6. Date
7. Time
8. Address of APPARATUS field:-
Isotope, collimator, scanner or camera
9. Address of ANATOMY field:-
Anatomic markers, view, orientation
10. Address of PROCESSING field:-
Smoothed, normalised, scale factor
11. Tag, for retrieval.
12. Spare.
13. Spare.
14. SCAN DATA.

Fig. 5. The structure of a scan file for use by the Varian Gamma camera users group.

In referring to archiving systems, one technique which may be of value is in the compression or encoding of scans so that the libraries may be correspondingly smaller, and retrieval correspondingly faster. There is considerable literature on this subject [5]. It may be added that there is some indication that the Hadamard transform may also have an application in this field since it can provide a reasonably efficient system of coding with a relatively rapid program. The design of the scan library itself and the parameters which may be used for retrieval is another topic of importance. It is our experience so far that as much information as possible should be stored, consistent with the difficulty of inputting this information into the computer. Thus, if for every scan it is necessary for the operator to complete a ten-page proforma then it will either not be done, or done incorrectly. However, a minimum should be the name and number of the patient, age, sex, isotope organ, view, date, time, apparatus (for example, collimator, sample size, etc.), plus a field for comments. Note that some of this, for example, date, time and apparatus, can be added automatically by the system itself. The system can and should add details of any manipulation, processing, etc., plus the anatomic reference data so that when the scan is recalled at a later date this information is still available and may be used for the comparison of one scan with another.

Also of interest in the field of image manipulations is the external definition of regions of interest. Unfortunately, such programs seem to be rather device-dependant at this stage, but programs for the definition of organ edges, and similar pattern recognition algorithms, must surely be developed, and exchanged. Likewise subjects of great potential interest are (rapid) tomographic reconstruction programs [6,7,8] on which some cooperation has already been possible, and programs for the correction of non-linearities.

Displays

The third of this tetralogy of subjects is that of display algorithms. Again the same problem of designing algorithms suitable for use on a small computer system exists. Fortunately if the display device used is a 'scope or an electrostatic plotter, it has been found that the complexity of the algorithms required for contour maps and isometric projection can be greatly simplified and such programs can occupy less than 2k bytes of a typical small computer. It has also been found that the kernel of such programs can be separated from the device drivers with the benefits that the same kernel can be used with several different displays, and that the same program can be used for different configurations, and therefore, users. The technique used at the Middlesex Hospital was to model all calls to the display device on the equivalent Calcomp routines, e.g. CALL PLOT (X,Y,PEN) where X and Y are the coordinates and PEN is an index originally used to determine whether the pen should be up or down, or the line dotted, etc. In this particular scheme the PEN index was used to determine whether the colour television, the electrostatic plotter, or the 611 was to receive the data, and to indicate in the first case the colour. It might be fruitful to incorporate a system such as this in the more

general N.M. language to which reference has previously been made.

The N.M. Language

This is rapidly becoming a subject of considerable importance in that every system of suitable size and complexity can only be used properly if there is a suitable system of program chaining, linking and overlaying. At the Middlesex Hospital, a system of this kind has been implemented. Individual modules have associated mnemonics. These mnemonics plus certain control data form the basis of a very simple language. New mnemonics can be defined in terms of old mnemonics. For example, the command BRAIN may be defined as equivalent to that series of commands needed to manipulate a brain scan. For reasonable flexibility it is necessary to include a command which is interpreted as "collect the next control data (or command) from a particular device (for example, the V.D.U.) or location in core". A typical statement in this language might be:

```
PROCESS = SMOOTH,VARIABLESMOOTH,$TELETYPE,5,3,NORMALISE.
```

where the command \$TELETYPE means collect this control information from the teletype, and SMOOTH, VARIABLESMOOTH and NORMALISE are three mnemonics. The ability to communicate with a location in core enables one module to modify another.

If, in addition, commands such as that discussed above for displays, plus commands similar to those being discussed for a CAMAC language [9], were to be added, then the basis of a more general N.M. language could be formed. The value of this would depend on whether its use would genuinely facilitate the exchange of algorithms and how much work would be entailed in its definition. It is to be suspected that this is rather futuristic, and that every centre will go its own way in defining such structures, all incompatible, but providing an occupation for some programmer or physicist.

Conclusions

A very brief summary of the field of nuclear medicine software has been given, together with a few examples of cases where cooperation has been possible, might be possible, or is needed. It seems likely that for this to be achieved, a deliberate effort must be made by those working in the field to develop their systems with such aims in mind. It is to be anticipated that the gains as a result of compatibility will outweigh the (slight) loss of efficiency.

BIBLIOGRAPHY

1. T. Gustaffson and A.E. Todd-Pokropek, "Filters with variable shape in radioisotope image processing", Proceedings of a symposium, Hannover 1971, The Data Processing and Data Handling of Radioisotope Scans, in press.
2. C. Berche, "Contribution a l'etude du filtrage numerique des images scintigraphiques", These de Medicine, Paris, 1972.
3. P. Pistor, "Digital processing of scintigraphic images by two-dimensional recursive Wiener filters", IBM Wissenschaftliches Zentrum, Heidelberg, Technical Report 70.12.006, December 1970.
4. P. Pistor, G. Walch, W. Ammann, P. Georgi, W.A. Hunt, W.J. Lorenz, H. Luig, H.G. Meder, P. Schmidlin, H. Wiebelt, "Digital Image Processing in Nuclear Medicine - Methods and Results", IBM Wissenschaftliches Zentrum Heidelberg, 71.05.002, May 1971.
5. H.C. Andrews, Computer Techniques in Image Processing, (Academic Press, New York-London, 1970).
6. A.E. Todd-Pokropek, "Two Methods for Improving Section Scanning", Proceedings of a symposium, Bad Gastein, 1972, Radioisotope Isotope in Klinik und Forschung, Band X (Urban and Schwarzenberg, Munich, 1972).
7. D.A. Chesler, "3-D activity distribution from multiple position scintigraphs", (abstract) J. Nucl. Med. 12, 347 (1971)
8. H. Tretiakoff and S. Goldstein, personal communication (March 1972).
9. P. Wilde, "A CAMAC Language", CAMAC Bulletin 3, 5-7 (March 1972).

CURRENT STUDIES WITH IDA (IMAGE DISPLAY AND ANALYSIS) SYSTEM ¹

H. N. Wagner, Jr., and T. K. Natarajan

ABSTRACT

The use of small general purpose computers in clinical nuclear medicine practice and research has increased steadily in the past few years. The Image Display and Analysis system we have developed satisfies many of the requirements of the field and additions or changes in programs can be made with ease. We are now using the IDA system daily in studies of the thyroid, lungs, cardiovascular system, hepatobiliary and renal systems and in radiopharmaceutical research and development. The system is proving versatile and very useful in (1) establishing objective quantitative criteria for the establishment of limits of normality of different organ functions; (2) determination of the efficacy of therapy; and (3) the monitoring of the natural history of a disease.

Division of Nuclear Medicine
The Johns Hopkins Medical Institution
Baltimore, Maryland

---

¹Supported by USPHS Grant GM 10548

CURRENT STUDIES WITH THE IDA (IMAGE DISPLAY AND ANALYSIS) SYSTEM

Henry N. Wagner, Jr. and T. K. Natarajan

Last year at the first symposium, we described our Image Display and Analysis system (IDA) for acquiring, processing, and analyzing radionuclide image data. We described some of the programs we have found useful. Since then, we have expanded the system and applied it widely in clinical Nuclear Medicine, as well as in research.

The present IDA system is shown in Figure 1. The major components (Figure 2) consist of:

1. a small general purpose computer with 32K of core memory;
2. a teletype;
3. a high speed paper tape reader;
4. two magnetic disc cartridges, 1660K words;
5. industry compatible (IBM) 9 track magnetic tape system;
6. a floating point processor giving dual processor capability;
7. a high speed multiplexer with priority assigned channels;
8. display subsystem consisting of a video monitor and mechanisms for beam steering and intensity modulation;
9. controls to vary image size and contrast;
10. a light pen for selection of areas of interest for further analysis and quantification.

Applications

Radiopharmaceutical Development

The IDA system has proved very useful in the field of radiopharmaceutical development to quantify the organ distribution and time course of radioactivity within the body in man and experimental animals.

Till now, the usual practice has been to administer the radiopharmaceutical to an animal, such as a mouse, remove the organs of interest, and count samples to determine the concentrations of the tracer in various regions. If the time course characteristics are of interest, many animals have to be sacrificed at different times after administration of the tracer.

The spatial and temporal resolution provided by IDA operating with a scintillation camera eliminates the need for sacrificing the animals, by allowing time course studies to be performed on the same animal. Even fast sampling times can be carried out with ease.

An example of such a study of an agent for studying gall-bladder function is shown in Figure 3. A series of images were recorded of the liver and gall-bladder region of a dog after the injection of a new radiopharmaceutical that

Figure 1: Image display and analysis (IDA) system

IDA IMAGE DISPLAY AND ANALYSIS SYSTEM

Figure 2: Block diagram of IDA System

5

20

60

120

180

240

Figure 3: Use of IDA to portray the passage of a new radiopharmaceutical through the liver and into the gall-bladder, seen as the progressively darkening region of activity in approximately the center of the liver of a dog. The numbers indicate time in minutes after intravenous injection of the tracer.

progressively accumulates in the gall-bladder. In the image 240 minutes after injection, the gall-bladder could be differentiated from the liver. Using the light pen, the gall-bladder was flagged in this image and the liver flagged in an earlier image. These flagged regions were then duplicated in all the serial images to obtain the quantitative time course of the tracer in the liver and the gall-bladder, as shown in Figure 4. After the gall-bladder activity has reached a plateau, a gall-bladder stimulating drug such as Cholecistakinine was injected to study the emptying function of gall-bladder. Such functional studies are designed to help in the differential diagnosis and treatment of gall-bladder disease.

THYROID

For the past year, we have been using the IDA system routinely for rapid evaluation of thyroid structure and function. The system is shown diagrammatically in Figure 5. With the pinhole collimator of the gamma camera positioned over the patient's thyroid in a way that the thyroid is magnified to fill the entire 10-in. diameter area of the scintillation camera, 5 mCi pertechnetate- ^{99m}Tc is injected intravenously. Data are accumulated in the core memory and are simultaneously and subsequently displayed as serial images on a video screen. The exact area and total and regional activity of the thyroid are computed by simple programs used together with a light pen. The extrathyroid activity is measured using the light pen to flag regions adjacent to the thyroid. The entire thyroid and areas of concern within the gland are characterized by rates of uptake of pertechnetate- ^{99m}Tc . The method requires only 10 minutes of the technologist's time and the results are available within 40 minutes of the start of the study. The patient need not return for subsequent study, since both structure and function are examined at the same time. Regional as well as total function can be measured.

Figure 6 illustrates typical differences among hyperthyroid, euthyroid and hypothyroid patients studied in this way. An example of regional measurement of thyroid function is shown in Figure 7, which portrays the images of the thyroid before and after the administration of triiodothyronine. There is suppression of normal thyroidal tissue, but an increase in uptake of tracer in the nodule after administration of triiodothyronine.

The computer is also used to assist in decision making (Figure 8). This program, an analysis of thyroid function, starts when the operator types the code letters on the teletype. The computer asks a series of questions, the answers to which (underlined) are typed by the operator. The operator has obtained most of the original data from the computer, which then calculates the net counts per minute, thyroid area, and percent thyroid uptake. Comparing these values against normal values stored in its memory, the computer determines the probability of abnormality. In the first case (Figure 8), the patient's thyroid function is probably normal, but in the second patient, the results indicate the probability of hyperthyroidism.

KIDNEY

Another example of the use of the light pen for selecting areas of interest for quantification is in the study of the kidney. The light pen, as opposed to fixed shape area cursors, is very flexible in flagging irregularly shaped areas within an organ.

Computer assisted interpretation of serial renal images obtained with the gamma camera has several major advantages: (1) asymmetry of function can be more readily detected (2) precise criteria for limits of normal variation can

Figure 4: Quantitative analysis of the data on which the images shown in Figure 3 are based. The progressive accumulation of the tracer in the gall bladder as it leaves the liver is readily measured.

Figure 5: Schematic diagram of the use of the image display and analysis (IDA) system for determining thyroid structure and regional function.

Figure 6: Typical computer-generated pertechnetate- ^{99m}Tc accumulation curves in hyperthyroid, euthyroid and hypothyroid patients.

Figure 7: Use of IDA for measurement of regional thyroid function in triiodothyronine (T_3) suppression test. Before administration of T_3 , the uptake of pertechnetate- ^{99m}Tc by the nodule was 0.4% of the administered dose per cm^2 at 20 minutes after administration of the pertechnetate dose, whereas that in adjacent thyroidal tissue was 0.1. After administration of T_3 the adjacent thyroid uptake fell by half, whereas the uptake in the nodule remained high, thus indicating the nodule was autonomous.

THYROID FUNCTION-TC

NAME: JOHN HIST NO: 453621 DATE: 29DEC71
 SYRINGE
 COUNTS: 40000 TIME: 540 RSDU: 4000 TIME: 620 PERIOD: 30
 PATIENT
 INJ TIME: 560 UPTAKE TIME: 660 PERIOD: 60
 AREA THYROID: 350 BKG: 100
 FACTOR: 98
 THY CNT: 10000 BKG CNT: 1000
 NET THY CPM: 6500.00
 THYROID AREA: 9.75 SQ CM
 THYROID UPTAKE: 1.15 % -- PROBABILITY: NORMAL --
 (1.151)

 MEAN NORMAL 0.90 %
 NORMAL 0.46 TO 1.76 % -1 TO +1 SD
 LOW NORMAL 0.46 TO 0.24 % -1 TO -2 SD
 HIGH NORMAL 1.76 TO 3.34 % +1 TO +2 SD
 HYP0- < 0.24 % -2 SD
 HYPER- > 3.34 % +2 SD

THYROID FUNCTION-TC

NAME: BROWN HIST NO: 126354 DATE: 28DEC71
 SYRINGE
 COUNTS: 40000 TIME: 540 RSDU: 4000 TIME: 600 PERIOD: 30
 PATIENT
 INJ TIME: 560 UPTAKE TIME: 660 PERIOD: 60
 AREA THYROID: 350 BKG: 100
 FACTOR: 88
 THY CNT: 50000 BKG CNT: 1000
 NET THY CPM: 46500.00
 THYROID AREA: 9.75 SQ CM
 THYROID UPTAKE: 8.25 % -- PROBABILITY: HYPERTHYROIDISM --
 (9.251)

 MEAN NORMAL 0.90 %
 NORMAL 0.46 TO 1.76 % -1 TO +1 SD
 LOW NORMAL 0.46 TO 0.24 % -1 TO -2 SD
 HIGH NORMAL 1.76 TO 3.34 % +1 TO +2 SD
 HYP0- < 0.24 % -2 SD
 HYPER- > 3.34 % +2 SD

Figure 8: Teletype output of thyroid uptake measurement

be established and (3) changes on sequential studies done in the same patient to assess the course of a disease or the efficacy of therapy can be determined.

In preliminary studies serial renal images are obtained for 30 minutes after 10-15 mCi i.v. dose of ^{99m}Tc DTPA. During the first minute, twenty consecutive 3 second exposures are obtained. Thereafter, the scan images are at consecutive 15 second intervals. The data are recorded as a 64 x 64 cell matrix per each serial frame on magnetic tape for subsequent analysis and display. Regions of interest flagged with a light pen, included right and left kidney, a strip of cortex (outer margin), the renal pelvis of each kidney, and the mid-abdominal aorta. The flagged regions are then transferred to all frames in the sequence to obtain the time course of activity in the various regions. Flagged areas below and lateral to the kidneys are used as indicators of background activity; background counts per unit area are subtracted frame by frame from counts per unit area are obtained in the regions of interest.

Quantitation with the technique may provide a sensitive means of identifying unilateral and asymmetric renal disease, and help to differentiate predominantly vascular from predominantly tubular injuries.

The use of the small general purpose computer is increasing more and more in clinical nuclear medicine and research. The image display and analysis system that we have developed over the past 4 years is now being used daily in studies of the thyroid, lungs, cardiovascular system, hepatobiliary, and renal systems.

Figure 9: Time course of ^{99m}Tc DTPA activity during the first minute after i.v. injection in a normal volunteer. The peak activity in both cortex sample and whole kidney occur very close to the aorta peak.

Figure 10: Time course of ^{99m}Tc DTPA activity during the first 20 minutes after i.v. injection in the same normal volunteer as Figure 9.

Figure 11: Time course of ^{99m}Tc DTPA activity during the first minute after i.v. injection in a patient with arterial stenosis. The peak activity of the cortex samples and whole kidneys are not as well defined as in Figure 9 and are delayed much further from the aorta peak.

Figure 12: Time course of ^{99m}Tc DTPA activity during the first 20 minutes after i.v. injection in the same patient as Figure 11.

REFERENCES

1. Kuhl, D. E., and Edwards, R. R.: "Digital techniques for on-site scan data processing" in Fundamental Problems in Scanning, edited by Alexander Gottschalk and Robert N. Beck (Springfield, Ill.: Charles C. Thomas).
2. Laughlin, J. S., Webber, D. A., Benua, R. S., Kenny, P. J., and Ritter, F.: "Quantitative storage, analysis and display in digital scanning" in Fundamental Problems in Scanning, edited by Alexander Gottschalk and Robert N. Beck (Springfield, Ill.: Charles C. Thomas).
3. Charleston, Donald B.: "An analog approach to scan readout data manipulation and enhancement" in Fundamental Problems in Scanning, edited by Alexander Gottschalk and Robert N. Beck (Springfield, Ill.: Charles C. Thomas).
4. Natarajan, T. K., and Wagner, H. N., Jr.: A new image display and analysis system (IDA) for radionuclide imaging, Radiology 93:4, 1969.
5. Tauxe, W. N., Chaapel, D. W., and Sprau, A. C.: Contrast enhancement of scanning procedures by high-speed digital computer, J. Nuclear Med. 7:647, 1966.
6. Harris, C. C., Bell, P. R., Francis, J. E., Jr. Jordan, J. C., and Satterfield, M. M.: "Data recording for radioisotope scanning", In Progress in Medical Radioisotope Scanning (US Atomic Energy Commission, TID 7673, 1962).
7. Kuhl, D. E., Chamberlain, R. H., Hale, J., and Gorson, R. O.: A high-contrast photographic recorder for scintillation counter scanning, Radiology 66: 730, 1956.
8. Kaihara, S., Natarajan, T. K., Maynard, C. D., and Wagner, H. N., Jr.: Construction of a function image from spatially localized rate constants obtained from serial camera and rectilinear scanner data. Radiology 93: 6, 1969.
9. Strauss, H. W., Natarajan, T. K., Sziklas, J. J., Poulou, K. P., Fukushima, T., and Wagner, H. N., Jr.: Computer assistance in the interpretation and quantification of lung scans, Radiology 97:2, 1970.
10. Natarajan, T. K., and Wagner, H. N., Jr.: "A data storage, processing and display system for radionuclide imaging" in Medical Radioisotopes Scintigraphy (Vienna: International Atomic Energy Agency, 1969.)
11. Proceedings of the symposium on sharing of computer programs and technology in nuclear medicine, Oak Ridge, Tennessee, April 1-3, 1971.
12. Wagner, H. N., and Natarajan, T. K.: "Computers in Nuclear Medicine," Year book of Nuclear Medicine, 1972.

A GENERAL LOGICAL STRUCTURE
FOR QUANTITATIVE ANALYSIS OF RADIOCARDIOGRAPHIC DATA

David M. Shames, M.D.

Nuclear Medicine Section
Department of Radiology
University of California, San Francisco, California

April 1972

A GENERAL LOGICAL STRUCTURE
FOR QUANTITATIVE ANALYSIS OF RADIOCARDIOGRAPHIC DATA

David M. Shames, M.D.
Nuclear Medicine Section
Department of Radiology
University of California, San Francisco, California

Abstract: Several different types of radiocardiographic data (RCD) obtained from the scintillation camera have proven useful in estimating physiological measures of the central circulation. These include high frequency data obtained over the region of the left ventricle for estimation of left ventricular ejection fraction, medium frequency data over multiple regions of the precordium and lungs for estimation of mean transit times of various chambers and cardiac output, and low frequency or equilibrium data gated by the electrocardiogram for estimation of left ventricular end-systolic and end-diastolic volumes. We have formulated a general logical structure within which all 3 types of RCD can be analyzed in an internally consistent fashion. This structure takes the form of a linear compartmental model which is fitted to the RCD by the SAAM-25 digital computer program. Data over a particular region are assumed to be linear combinations of count rates from overlapping chambers. Although the model is somewhat loose in the statistical sense due to high covariance between the linear coefficients and actual fractional turnover rates of the model, uniqueness is possible when data from more than one kind of injection are analyzed together. Steady state solution of the model yields estimates of all chamber volumes and all forward and backward flows including shunt and valvular regurgitant flows. The covariance matrix obtained from the least squares fit yields uncertainty estimates of the steady state measures. The technique is noninvasive and the logical machinery for doing the data analysis is completely contained within the relatively easy to use SAAM-25 program.

Several different types of radiocardiographic data obtained from the scintillation camera interfaced with a digital data acquisition system have proven useful in estimating clinically important physiological measures of the central circulation. Examples of these data, referred to as high, medium, and low frequency data, are shown in Fig. 1. High frequency data, obtained over the region of the left ventricle at a rate of 10-20 frames per second, can be used to estimate left ventricular ejection fraction (LVEF) following subtraction of time-dependent background activity^{1,2}. The LVEF is calculated by taking the ratio of the difference between any peak and succeeding through count rates to that peak count rate. Medium frequency data are those obtained at a rate of 1 to 2 frames per second over various regions of the precordium and lungs. These data, often referred to as isotope dilution curves, can be used to estimate mean transit times through various chambers of the heart and lungs and cardiac output. When combined, these measures yield estimates of cardiac chamber volumes, pulmonary blood volume and forward and backward flow rates in the central circulation³. Low frequency data are those obtained over

THREE TYPES OF RADIOCARDIOGRAPHIC DATA

I. HIGH FREQUENCY DATA (10 - 30 frames/sec)

II. MEDIUM FREQUENCY DATA (1 frame/sec)

III. LOW FREQUENCY DATA ("equilibrium data")

Figure 1

various regions of the precordium following circulatory mixing of an intravascular label such as ^{99m}Tc -albumin. These data are required to calibrate the medium frequency data over the precordium for cardiac output determination. When gated by the electrocardiogram, these low frequency data can also be used to estimate left ventricular end-systolic volume, left ventricular end-diastolic volume and LVEF⁴.

All three types of radiocardiographic data can be analyzed at the same time in terms of the general logical structure shown in Fig. 2. This structure takes the form of a linear compartmental in which various parts of the central circulation are represented by compartments. The fractional turnover rates of these compartments λ_i , are the reciprocals of the mean transit times through the corresponding heart chambers. Forward as well as backward flow rates corresponding to shunt and valvular regurgitant flows are described. The model is fitted primarily to the medium frequency data obtained over the cardiac chambers described in the model. The initial condition for the tracer solution is either a pulse directly into the right atrium by catheter or a pulse into a peripheral vein. The magnitude of the pulse is equal to the amount of administered activity. In the case of injection into a peripheral vein, the time function over the superior vena cava is monitored and becomes the tracer input function of the model. Fitting the model to the precordial data subject to this superior vena cava input function yields a deconvolution of the precordial data from the input function. In this way the transit kinetics of the tracer between an antecubital vein and right atrium are not confused with the transit kinetics of the tracer through the central circulation.

Least squares fitting of the constants of the model to all the medium frequency data as well as calculation of cardiac output is performed using the Berman-Weiss N.I.H. SAAM-25 digital computer program operated on a Univac 1108 computer⁵ Tracer solution of the model yields estimates of all fractional turnover rates, a set of proportionality constants, K_i , relating external count rate to actual activity in the various cardiac chambers and cardiac output. Steady state (tracee) solution of the model (also performed by SAAM at the same time as the tracer solution) using the calculation of cardiac output as input vector, yields estimates of all chamber volumes, pulmonary blood volume and all forward and backward flow rates. The covariance matrix obtained from the least squares solution yields uncertainty estimates for all volumes, flows and fractional turnover rates.

Other important information obtained in the study, such as LVEF from the high frequency data and left ventricular end-diastolic volume (LVEDV) from the low frequency data are entered into the data stream as independent statistical constraints on the tracer and steady state solutions respectively. As shown in Fig. 2, the total fractional turnover rate of the left ventricle (backward and forward) can be estimated from the product of the LVEF (obtained from the high frequency data) and heart rate. This value is entered as the initial estimate of the total fractional turnover rate of the left ventricle along with any desired standard deviation. The estimate of LVEDV is also entered as an initial estimate of the steady state calculation of LVEDV again with any desired standard deviation. In this way the tracer and steady state solutions obtained from the medium frequency data are fitted to these data subject to the additional statistical constraints that the LVEF and LVEDV are known independently within certain limits.

While the model in Fig. 2 has appeal because of its mathematical simplicity and potential clinical usefulness, it does not adequately describe the reality of external count rates over various regions of the precordium. These data, in fact, represent linear combinations of count rates from more than one cardiac chamber and usually lung. A model describing these linear combinations of count rates is shown in Fig. 3 where the linear coefficients are represented

CENTRAL CIRCULATION MODEL

MEDIUM FREQUENCY DATA

HIGH FREQUENCY DATA

$$(LVEF) \times (HR) = (\lambda_{LV-Ao} + \lambda_{LV-MI})$$

SOLUTIONS

TRACER $\rightarrow \lambda, K, CO$

TRACEE $\rightarrow$ all volumes and flows

LOW FREQUENCY DATA

LVEDV, LVEF

Figure 2 - Circles represent cardiac chambers and lung. Solid arrows represent forward and backward fractional turnover rates respectively. Small triangles overlying circles represent proportionality constants, K_i , relating external count rate to actual amount of activity in a given chamber. Abbreviations include SVC for superior vena cava, RA for right atrium, RV for right ventricle, L for lung, LA for left atrium, LV for left ventricle, Ao for aorta, MI for mitral insufficiency, ASD for atrial septal defect, VSD for ventricular septal defect, λ_i for fractional turnover rate of compartment i , LVEF for left ventricular ejection fraction, HR for heart rate, LVEDV for left ventricular end-diastolic volume and CO for cardiac output.

CENTRAL CIRCULATION MODEL

MEDIUM FREQUENCY DATA

HIGH FREQUENCY DATA

$$(LVEF) \times (HR) = \lambda LV$$

LOW FREQUENCY DATA

1. LVEDV, LVEF
2. INDEPENDENT CALCULATION OF CHAMBER VOLUMES

$$R_i(e) = C_e V_i K_i$$

$$V_i = \frac{R_i(e)}{C_e K_i}$$

SOLUTIONS

TRACER $\rightarrow \lambda, K, CO$

TRACEE $\rightarrow$ all volumes and flows

Figure 3 - Large triangles and dotted lines represent linear combinations of count rates. Abbreviations are the same as those already defined in Figure 2. Additional abbreviations include LAO for left anterior oblique, RAO for right anterior oblique, $R_i(e)$ for external count rate over chamber i at equilibrium, C_e for activity concentration in blood at equilibrium and V_i for volume of chamber i .

by dotted lines. The large triangles represent precordial count rates over the left side of the heart which are assumed to be linear combinations of individual count rates from various cardiac chambers. Here a count rate over the left ventricle in the left anterior oblique projection represents a combination of count rates from the left ventricle, left lung and perhaps descending aorta. Similarly, count rates over the left atrium and aorta also represent combinations of count rates. Unfortunately the high covariance between the linear coefficients and actual fractional turnover rates of this model render the model non-unique with respect to all the medium frequency data and the additional statistical constraints previously mentioned. This implies that the parameters of the model may have an infinite number of physically realizable values all of which fit the data with maximum likelihood.

Additional data are required to render the model unique. These might include a repeat injection in another view in which case the fractional turnover rates of the model would remain the same while the values of the linear coefficients would change. Figure 3 shows that in the right anterior oblique view, (as opposed to the left anterior oblique view), a count rate over the left ventricle would represent a combination of count rates from the left ventricle, right ventricle and lung. Additional information with respect to cardiac chamber volumes can be extracted from the low frequency data obtained over the same region of the precordium from which the medium frequency data were obtained. This relation is shown in Fig. 3 under the heading "low frequency data". A set of low frequency count rates, $R_{ij(e)}$, obtained over given regions of the precordium, when combined with the equilibrium activity concentration in the blood, C_e , and the set of K_i obtained from the medium frequency data over the same regions of the precordium, yield a set of chamber volumes, V_i . These volume estimates must be equal to those volume estimates obtained from the tracee solution derived by applying the cardiac output as input vector to the set of fractional turnover rates, λ_i . Similar logic is used to relate the calculation of a linear combination of volumes from the low frequency data to the independent calculation of the same linear combination of volumes from the medium frequency data.

Even with all these constraints, the model shown in Fig. 3 remains somewhat loose in the statistical sense due to the high covariance between the linear coefficients of the model and fractional turnover rates. At the present time we are looking into the value of other constraints such as an independent estimate of the fraction of counts coming from the lung over any given region of the precordium and more direct estimates of the kinetic parameters of the left side of the heart such as might be obtained from an injection into a segmental pulmonary artery.

REFERENCES

1. D. Van Dyke, H.O. Anger, R.W. Sullivan, Y. Yano and H. Parker, "Cardiac Evaluation from Radioisotope Dynamics," J. Nuc. Med. in press.
2. P. Weber, L.V. dos Remedios and I.A. Jasko, "Quantitative Radioisotopic Angiocardiology." Submitted for publication.
3. Y. Ishii and W.J. MacIntyre, "Measurement of Heart Chamber Volumes by Analysis of Dilution Curves Simultaneously Recorded by Scintillation Camera," Circulation 44, 37-46 (July 1971).
4. H.W. Strauss, P.J. Hurley, B. Pitt and H.N. Wagner, Jr., "Scintiphotographic Method for Measuring Left Ventricular Ejection Fraction in Man without Cardiac Catheterization," Am. J. Cardiol. 28, 575-580 (1971).
5. M. Berman and M.F. Weiss. Users' Manual for SAAM. U.S.P.H.S. Publication No. 1703. U.S. Government Printing Office, Washington, D.C.

LOCALIZATION OF HEART CHAMBERS BY TEMPORAL AND SPATIAL ANALYSIS
FOR REGIONAL MYOCARDIAL BLOOD FLOW MEASUREMENTS

William J. MacIntyre, Robert E. Botti, Yasushi Ishii, and Thomas S. Houser

Department of Medicine, Case Western Reserve University
and University Hospitals, Cleveland, Ohio 44106

ABSTRACTS

When ^{43}K is injected into a peripheral vein the relative deposition of the material in the myocardium will be dependent upon the blood supply perfusing this region and consequently will reflect the relative myocardial blood flow to that region. The uptake of radiopotassium is not selectively confined to the myocardium, however, and thus myocardial deposition is difficult to differentiate from the amounts taken up by the liver and other adjacent organs and tissue. This differentiation or delineation of the heart border has been accomplished by recording by scintillation camera the rapid passage of a bolus of $^{99\text{m}}\text{Tc}$ as it passes through the heart chambers and superimposing the heart border so determined over the distribution recorded by the accumulation of ^{43}K over a longer period.

April 1972

This investigation has been supported by the National Heart and Lung Institute Grant HE-06304, National Institutes of Health, U. S. Public Health Service

INTRODUCTION

The measurement of myocardial blood flow by radionuclide techniques has been accomplished primarily by three general methods: one, static techniques which measure the amount of a non-metabolized material trapped in the capillaries of the myocardium ^{1,2}; two, dynamic techniques in which the rate of accumulation ^{3,4} or the rate of washout ^{5,6} of a material is measured; and three, the fractionation method ^{7,8} in which the fraction of the cardiac output perfusing the myocardium is determined.

In the static technique, particulate matter in such form as macro-aggregated serum albumin or albumin microspheres tagged with either ¹³¹I or ^{99m}Tc is commonly used. Since the halftime of clearance of these materials from the myocardium is in the range of a few hours, visualization of the relative deposition has been successfully accomplished by both moving detector scanners and by scintillation cameras.

The washout technique involves perfusion of the myocardium by an inert diffusible tracer which will subsequently be cleared from the tissue by blood flow. The noble gases provide the best example of this, with ¹³³Xenon being the most common of the radionuclides used for this purpose. Because of the rapid removal of the radioactive material both single probes and dynamic camera recording have been found most useful for this measurement.

In both of the above techniques the radioactive material is introduced directly into the coronary circulation. Delineation of the heart chambers, therefore, is relatively simple since the regions supplied by the specific coronary arteries are well known. The primary disadvantage of these methods lies in the traumatic procedure by which the radionuclide material must be given, i.e. injection into a coronary artery. For that reason attention has been directed to the possibilities of similar information derived by the less traumatic procedure of injecting into a peripheral vein.

These peripheral vein injection methods involve either measurement of the accumulation rate of a material by the myocardium, or measurement of the fractional cardiac output. Both of these procedures have utilized either radiopotassium or a potassium analog such as rubidium for which the myocardium exhibits a relatively high extraction efficiency. The difficulty of this measurement is that other organs and tissues also extract these substances so that the border of the heart may be indistinct or actually overshadowed by adjacent organs such, as the liver, which may exhibit a higher total uptake than the myocardium ⁹.

It has been demonstrated that individual heart chambers can be identified from observations made during the rapid passage of a bolus injection through the heart ^{10,11,12}. Therefore it was considered feasible to superimpose the heart border so determined over the precordial scan which records the entire potassium deposition in the heart. In this way regions not involving the heart can be eliminated and in addition the magnitude of deposition can be related to specific regions of the myocardium. This present report is concerned with the methodology involved with the delineation of the heart chambers by the rapid passage of radioactive material through the central circulation and its use in determining regional relative myocardial blood flow.

MATERIAL AND METHODS

Single and twin probe detection of the fraction of injected dose deposited in the myocardium has been confined largely to measurements with ⁴²K and ⁸⁶Rb. These radionuclides are not adaptable to scintillation camera use, however, because of the high energy gamma emission. The availability of ⁴³K with its 371 Kev gamma ray in 85% of its disintegrations has made camera visualization of the myocardium now practical ¹³.

At this energy the conventional 3 inch collimator has been unsatisfactory because of septal penetration; thus a 4 1/2 inch collimator has been preferred for these measurements. Even with the longer collimator considerable septal

penetration has been observed. The spread function of the scintillation camera for a line source of ^{43}K is shown in Figure 1. Because of the long tails, deconvolution of the image has thus been required for quantitative results 14,15.

The patients were studied in the supine position with a Nuclear Chicago Eco-Gamma II scintillation camera placed over the precordium inclined to approximately 30° in the left anterior oblique position so that the camera face was perpendicular to the plane of the septum, thus separating the right heart from the left. An additional inclination of 15° along the longitudinal axis of the patient was utilized to aid in separation of the atria from the ventricles.

With the patient in place under the camera a dose of ^{43}K Cl varying between 0.4 and 1.0 millicuries in a volume of 3 ml was rapidly injected into an antecubital vein followed by 10 ml of saline flush solution. The distribution of the material during the rapidly changing sequence was recorded every 0.6 seconds by an RIDL 1600 channel analyzer as a digitized 40×40 matrix as previously described 16,17. Each frame was transferred to an Ampex TM-7291 digital tape recorder for processing by a GA 18/30 computer.

At the end of one minute the frame accumulation time was increased to 60 seconds and recorded at this rate for the next 18 minutes. Following the recording of the last ^{43}K distribution, the spectrometer of the scintillation camera was changed to the $^{99\text{m}}\text{Tc}$ setting and a background count of the residual ^{43}K falling in the $^{99\text{m}}\text{Tc}$ window was recorded. A rapid injection of 5 to 8 millicuries of $^{99\text{m}}\text{Tc}$ was then made by similar flushing techniques. The rapid phase was again recorded every 0.6 seconds until the monitor ratemeter showed the material clearing from the heart.

APPLICATION OF TECHNIQUES

A sequence of print-outs of individual 40×40 matrices during the rapid passage of the $^{99\text{m}}\text{Tc}$ through the central circulation is shown in Figure 2. Various symbols were printed out to represent the contours at cut-off levels of 25%, 40%, 60% and 80%. In frame 3 at the top left the $^{99\text{m}}\text{Tc}$ pertechnetate has just arrived at the superior vena cava. In frame 4 the injected material has progressed into the chambers of the right heart and by frame 5 some activity at the upper left shows a progression of the radioactivity from the right ventricle out to the pulmonary artery. Each frame represents a collection time of 0.6 seconds and it requires approximately 0.3 seconds to transfer the RIDL 1600 memory to digital tape for a total of 0.9 seconds between frames.

By frame 8 the pertechnetate is fairly well distributed in the lung circulation and frames 12 and 14 show the return of the radioactive material from the pulmonary circulation back to the chambers of the left heart. Note in frame 12 that there is some residual radioactivity still clearing from the lung circulation.

Frames 3,4 and 5 can be lumped together as shown in Figure 3 so that the composite represents the right side of the heart. The actual counts accumulated during these three frames is shown by the matrix on the upper left and the symbols depicting the various level contours are shown on the lower left.

The left side of the heart can be similarly delineated by lumping frames 11,12,13,14 and 15. Inclusion of frames after this point would reflect radioactivity in the ascending and descending aorta, while distributions earlier than frame 11 would be confusing because of including some radioactive material present in the pulmonary circulation.

Finally a composite of both sides of the heart is illustrated in Figure 4. Again the actual number of counts collected in each element is shown on the top and the contour symbols on the bottom. Note that the symbols denoting the 25% and the 40% contour levels are close together, usually no more than one element apart, demonstrating that the fall off of the edge of the heart border is quite steep.

The three dimensional views of the matrices in Figures 3 and 4 are shown

Figure 1. Line spread function of scintillation camera with 4 1/2 inch 1000 hole collimator, measured with ^{43}K .

Figure 2. Composite of six contour symbol matrices reflecting the passage of ^{99m}Tc through the central circulating system. Top left frame shows initial appearance of radioactivity in the superior vena cava. Middle and bottom frames on the left are at 0.9 second intervals following initial appearance. Top right frame shows distribution in lung circulation at 2.7 seconds later and following two frames show return to the left chambers of the heart.

Figure 3. Numerical matrices and contour symbols show right (top) and left (bottom) chambers of the heart.

Y COORDINATES ARE ON HORIZONTAL AXIS. X COORDINATES ARE ON VERTICAL AXIS

SYMBOLS REPRESENT CONTOURS AT LEVEL LISTED

- = 10 (20.0 3)
- = 20 (40.0 3)
- = 30 (60.0 3)
- = 40 (80.0 3)

259

in Figure 5. The upper figures represent the right heart with lumped frames 3 through 5, the middle view shows the lumped frames 11 through 15, and the lower view shows the summation of the two. The two views illustrated are taken from the heart with left and right projections. Two similar projections from the apex of the heart looking towards the base are also recorded but are not shown here.

The three dimensional models in Figure 5 are shown with all information recorded. As a result, activity in the lung circulation is included and in many cases activity in the descending aorta is observed. By recording only those elements of the matrix whose magnitude is at least 30% of the peak count, however, these extraneous areas can be eliminated. The three dimensional views of Figure 6 shows this elimination in which only those regions of Figure 5 which exceed the contour line of 30% are displayed.

The selected contour lines can now be applied to the full field deposition of ^{43}K in order to determine what part of the field is occupied by the heart. A second example in Fig 7 shows the precordial measurement of ^{43}K deposition in the chest recorded from 3 to 15 minutes following injection. While the heart exhibits the highest concentration, the uptake of other regions is considerable. The top model shows the accumulated counts of the rapid phase frames 7 and 8 and 13 through 17. A 30% contour is derived from the top model and the center figure shows the portion of the lower figure falling within the 30% contour line. This figure then represents the relative deposition of ^{43}K in the myocardium, uncorrected for amount of radioactivity remaining in the blood or deposited in muscle tissue anterior or posterior to the heart.

In the lower model of Figure 8 the heart is seen to be quite prominent with a significant displacement in the central portion as recorded from a nine minute accumulation of ^{43}K deposition. By applying the contour line from the $^{99\text{mTc}}$ rapid phase passage as shown in the top figure, the boundary of the heart can be delineated as shown by the middle figure.

It is uncertain whether the central displacement is occurring in the right heart, left heart or between the two. This information can be obtained by delineating only that region of the slow phase that falls within the border of the right heart chambers shown on the rapid $^{99\text{mTc}}$ frames. The upper model on Figure 9 shows the right heart as observed during frames 6,7 and 8 of the $^{99\text{mTc}}$ passage. The middle figure shows the ^{43}K deposition falling within the 30% contour of the rapid phase. The deposition is seen to be to the right of the region containing the central defect of Figure 8 and indicates that the displacement seen in Figure 8 must be entirely within the left ventricle.

DISCUSSION

Previous work on the deposition of ^{43}K in the myocardium of the dog ¹⁴ had indicated that the heart borders were relatively easy to determine except for the proximity of the liver to the lower border (Figure 10). This lower border of the heart could be identified quite easily by observation of the rapid passage of the ^{43}K prior to recording the deposition in the myocardium.

Extension to patient measurements has proved somewhat more complicated. For this reason a separate injection of $^{99\text{mTc}}$ for heart chamber identification alone has been performed to obtain greater statistical accuracy during the rapid phase.

The selection of the 30% contour on the cut-off level is arbitrary and is not necessarily the same value for every subject. In some cases where the cardiac output is very low the clearance from the lung is slow and may obscure the left border of the heart. In those cases only the right side may be definitive and the left side may then be normalized to its own maximum value for a better determination of relative deposition.

In the animal experiments the deposition of ^{43}K in the entire heart outline was summated to determine total myocardial flow. It is feasible to obtain the same determination in patients but of even greater interest is the

205229 305 SCALE= 34 MMX = 34 FRAMES 2 THRU 5. 99% CONTOUR

205229 1115 SCALE= 34 MMX = 31 FRAMES 11 THRU 15. 99% CONTOUR

205229 715 SCALE= 34 MMX = 52 FRAMES 3-5, 11-15 . 99% CONTOUR

TC99M HEART STUDY, RAPID PHASE

Figure 5. Three dimensional representation of the matrices in Figure 3 (top and middle) and Figure 4 (bottom).

206289 305 SCALE= 30 NMAX = 46 RAPID PHASE LUMPED FRAMES 3 THRU 5. AVS

206289 1115 SCALE= 34 NMAX = 52 RAPID PHASE LUMPED FRAMES 11 THRU 15. AVS

206289 315 SCALE= 35 NMAX = 54 RAPID PHASE LUMPED FRAMES 3-S.11-15. AVS

206289 315 0.6 0.28H

TC99M

HEART STUDY, RAPID PHASE

Figure 6. Three dimensional representation similar to Figure 5 but plotting only those values 30% or greater than the peak value. This discrimination removes most regions of lung circulation.

LUMPED RAPID FR. 7-8. 13-17 CONTOUR AT 30%

DATA WITHIN 30% CONTOUR, S-ELEM. AVERAGE

DECONVOLVED (1,1) DATA, S-ELEM. AVERAGE

K43 LUMPED SLOW PHASE DATA.

FRAMES 23 THRU 37

Figure 7. Three dimensional representation of the lumped frames of rapid passage of ^{99m}Tc (top), the deposition of ^{43}K in the chest region (bottom), and the deposition of the precordial distribution falling within the borders of the heart as determined by determining the 30% contour level of the top model and displaying only the portion of the bottom figure falling within that contour level. This is a different patient from the previous example, note the selection of different frames for this subject.

RAPID PHASE LUMPED DATA, CONTOUR AT 35%

SLOW PHASE DATA LYING WITHIN CONTOUR, AVS

BASIC SLOW PHASE LUMPED DATA, AVS

Figure 8. Distribution of ^{43}K in patient showing marked displacement in central area of myocardium. Top view represents $^{99\text{m}}\text{Tc}$ distribution, middle view shows portion of bottom figure falling within 30% contour level.

RAPID PHASE LUMPED DATA, CONTOUR AT 30%

SLOW PHASE DATA LYING WITHIN CONTOUR, AVS

BASIC SLOW PHASE LUMPED DATA, AVS

Figure 9. Same patient as Figure 8 but showing only the passage of ^{99m}Tc through the right side of the heart and applying the contour line of that portion to the ^{43}K distribution. Middle figure then represents the ^{43}K deposited in the right heart only.

FROM

LUMPED DATA

Figure 10. Three dimensional view of distribution of ^{43}K in the dog measured in the left anterior oblique position. Note reasonable isolation of heart but greater deposition in the liver.

ability to determine relative flow by a procedure involving only peripheral vein injection.

If this technique proves useful in the assessment of regional myocardial flow, a large number of clinical evaluations could be undertaken with this methodology. In addition to the relationship of time to the change in effective myocardial flow following infarction, the effect of vasodilation and the effect of coronary vein by-pass grafts would be of primary interest. Since the procedure is relatively non-traumatic, its usefulness as a screening test for changes in regional myocardial flow with age, exercise, diet or other factors would merit serious consideration.

ACKNOWLEDGMENTS

The computer facilities were provided by the George M. Humphrey Cardio-pulmonary Laboratory of University Hospitals of Cleveland. Miss Elaine A. Roth and Mr. George Roush have provided great assistance in the measurements on patients. The ^{43}K has been made available by the efforts of Dr. J. K. Poggenburg of the Isotope Development Center of Oak Ridge National Laboratories.

The original concept of this project has been stimulated greatly by many discussions with Dr. Walter H. Pritchard.

REFERENCES

1. W.L. Ashburn, E. Braunwald, A.L. Simon, K.L. Peterson, and J.H. Gault, "Myocardial Perfusion Imaging with Radioactive-labeled Particles Injected Directly into the Coronary Circulation of Patients with Coronary Artery Disease," Circulation, 44, 851 (1971).
2. M. Endo, T. Yamazaki, S. Konno, H. Hiratsuka, J. Akimoto, T. Tanaka, and S. Sakakibara, "The Direct Diagnosis of Human Myocardial Ischemia Using ^{131}I -MAA Via the Selective Coronary Catheter," Am. Heart J., 80, 498 (1970).
3. W.D. Love, and G.E. Burch, "A Study in Dogs of Methods Suitable for Estimating the Rate of Myocardial Uptake of Rb^{86} in Man, and the Effect of L-Norepinephrine and Pitressin on Rb^{86} Uptake," J. Clin. Invest. 36, 468 (1957).
4. R.J. Bing, A. Bennis, G. Bluemchen, A. Cohen, J.P. Gallagher, and E.J. Zaleski, "The Determination of Coronary Flow Equivalent with Coincident Counting Technic," Circulation, 29, 833 (1964).
5. R.S. Ross, K. Ueda, P.R. Lichtlen, and J.R. Rees, "Measurement of Myocardial Blood Flow in Animals and Man by Selective Injection of Radioactive Inert Gas into the Coronary Arteries," Circulation Res., 15, 28 (1964).
6. P.J. Cannon, J.I. Haft, and P.M. Johnson, "Visual Assessment of Regional Myocardial Perfusion Utilizing Radioactive Xenon and Scintillation Photography," Circulation, 40, 277 (1969).
7. L.A. Sapirstein, "Fractionation of the Cardiac Output in Rats with Isotopic Potassium," Circulation Res., 4, 689 (1956).
8. L.A. Sapirstein, "Regional Blood Flow by Fractional Distribution of Indicators," Amer. J. Physiol., 193, 161 (1958).
9. W.D. Love, R.O. Smith, and P.E. Pulley, "Mapping Myocardial Mass and Regional Coronary Blood Flow by External Monitoring of ^{42}K or ^{86}Rb Clearance," J. Nucl. Med., 10, 702 (1969).
10. D.T. Mason, W.L. Ashburn, J.C. Charbert, L.S. Cohen, and E. Braunwald, "Rapid Sequential Visualization of the Heart and Great Vessels in Man Using the Wide-Field Anger Scintillation Camera," Circulation, 39, 19 (1969).
11. J.P. Kriss, "Diagnosis of Pericardial Effusion by Radioisotopic Angiocardiology," J. Nucl. Med., 10, 233 (1969).
12. P.J. Hurley, H.W. Strauss, and H.N. Wagner, Jr., "Radionuclide Angiocardiology in Cyanotic Congenital Heart Disease," The Johns Hopkins Med. J., 127, 46 (1970).
13. M. Cooper, P. Hurley, and R. Reba, " ^{43}K as a Myocardial Scanning Agent," J. Nucl. Med., 11, 310 (1970).
14. W.J. MacIntyre, Y. Ishii, W.H. Pritchard, and R.W. Eckstein, "Measurement of Total Myocardial Blood Flow with ^{43}K and the Scintillation Camera. Myocardial Blood Flow in Man" (in press). Proceedings of the Pisa Symposium, June 1971.
15. T. Nagai, T.A. Iinuma, and S. Koda, "Computer-Focusing for Area Scans," J. Nucl. Med., 9, 507 (1968).

16. Y. Ishii and W.J. MacIntyre, "Measurement of Heart Chamber Volumes by Analysis of Dilution Curves Simultaneously Recorded by Scintillation Camera," Circulation, 44, 37 (1971).
17. W.J. MacIntyre, S.R. Inkley, E. Roth, W.P. Drescher, and Y. Ishii, "Spatial Recording of Disappearance Constants of Xenon-133 Washout from the Lung," J. Lab. Clin. Med., 76, 701 (1970).

COMPUTER MEDICINE, PHASE II: FROM INNOVATORS TO IMPLEMENTERS

Keith W. Sehnert, M.D.

Society for Computer Medicine

ABSTRACT

Computer Medicine: Phase I was the era of conceptual computer work for medical use. It was done by innovators in medical schools, private and Federal research projects. Computer Medicine: Phase II began with the decade of the 70's when a rapid diffusion of these concepts began to go out from medical centers to middle-sized and small medical centers and the offices of individual physicians. This diffusion is being aided by the implementers who are applying practical and cost effective methods to computer applications for widespread patient care. The most common applications are being seen in these five areas: ECG analysis, laboratory data systems, nuclear medicine and X-ray diagnostics, monitoring and I.C.U. systems, and medical record systems.

April 1972

As I was preparing my thoughts for this presentation and wondering about my topic, Computer Medicine: Phase II, one of my colleagues in the Society for Computer Medicine said, "I don't see how you can describe Computer Medicine at all, much less divide it into phases. I don't think we know where we are."

That comment reminded me of a recent flight I took from Dulles Airport to Chicago. The weather was bad and after we had been airborne for about an hour the pilot's voice came over the intercom with this message: "Good evening, folks. I've got some news for you. Some bad - some good. First, the bad, I don't know for sure where we are. Now the good, but we're 25 minutes ahead of schedule!"

Well, I think that might be a good description of Computer Medicine for many observers, but I think I know where we are and perhaps after you hear my presentation, I hope this audience will think I'm a better navigator than the Washington-Chicago pilot.

As the Editor of Computer Medicine each month I screen over 100 news items: newspaper clippings, medical papers and various other pieces of literature that relate to some type of medical and hospital computer activity in the United States. When my publication was started in late 1970, the monthly flow averaged 30-40 items. When I first started reviewing the field in the late Sixties, I doubt if I could have found enough medical computer news to fill one issue.

Now, in addition to a monthly publication about these events, there is a fledgling group called the Society for Computer Medicine. This group of nearly 150 physicians, hospital administrators, engineers, computer specialists, nurses and other health care professionals is devoted to bringing advancements of computer technology as rapidly as possible into the medical world.

What happened as the Decade of the Seventies opened its doors? I believe that "Computer Medicine: Phase II" was ushered in.

"Computer Medicine: Phase I" was the era of conceptual medical computer applications. It involved the traditional R & D work by medical schools, private and federal research projects. The Innovators were at work. They developed methods and ideas. They started things and predicted what the future would bring.

As is normal, following the initial creative idea, in medicine or science, the early efforts of improvement and application are often met with frustration.

A pioneer¹ in the specialty of computer medicine, G. Octo Barnett, M. D., of Harvard Medical School, said in 1970, "For the past decade it has been repeatedly claimed that computers will be of enormous usefulness in patient care and hospital practice. On innumerable occasions, in hundreds of articles

and editorials, in dozens of symposiums, our old men have dreamed dreams and our young men seen visions.

Yet, when I critically examine what computer applications are working, on an operational, continuing service basis in patient care activities, I am most impressed by the number of projects which have either been abandoned or in which the original grandiose objectives have been greatly diluted. This state of underachievement may be confusing to a casual observer, since the need seems so obvious and the technology that has been demonstrated in other fields to be so powerful."

In his book, Future Shock, Alvin Toffler² said, "Technological innovation consists of three stages, linked together into a self-reinforcing cycle. First, there is the creative, feasible idea. Second, its practical application. Third, its diffusion through society.

The process is completed, the loop closed when the diffusion of technology embodying the new idea, in turn, helps generate new creative ideas. Today there is evidence that the time between each of the steps in this cycle has been shortened."

Many of us are aware of events associated with milestones in medical care and the time frames involved. An example that comes to my mind occurred at my alma mater, Case-Western Reserve, where Dr. Claude Beck, in 1938, for the first time in annals of medicine, attempted to defibrillate a human heart in an effort to restore the patient's heartbeat.

Although this patient died later from brain damage, a feasible idea was born. It was not until 1947 that Dr. Beck performed the first successful defibrillation of the heart - and the 14-year-old patient went on to recover fully from his illness and become a healthy adult. I remember watching similar dramatic events when I was in medical school in Cleveland in the early 50's.

By 1960 the technic had been further improved and the practical application was being extended to other major medical centers.

Now in 1972 the technic has diffused through our society and is used daily around the world with great medical benefit. The time-frame was 30 years.

It took 169 years between the development of the microscope by Leewenhock in 1683 and 1852 when Virordt used it to count blood cells.

The time-frame between Auenbrugger's idea of percussion in 1761 until Laennec's stethoscope was put into clinical use in 1864 was 103 years.

Now in jumping to present times, the computer was invented only 25 years ago at the University of Pennsylvania by Drs. Eckert and Mauchly - and look at its rapid development and diffusion into society. A brief time-frame.

The initial programs for automated electrocardiographic systems developed by Caceres, then with USPHS in Washington, Fordy at Mt. Sinai Hospital in New York and Smith and Hyde at the Mayo Clinic were started in the early 60's. Some of the systems were operational in 1966. Now by 1972 computerized ECG analysis is being used in communities all around the country. A time-frame of only ten years!

As I indicated earlier, the medical and hospital literature of Phase I described what was being done by the Innovators at certain medical and research centers. The articles presented methods being used in the Fifties and Sixties and predicted what would happen in the future. A typical proposal, the

hospital information systems concept, will be a reality in about 100 hospitals by 1975.

Two surveys that appeared in 1971 showed that these techniques and applications were being transplanted to a substantial number of smaller hospitals, medium-sized centers and medical clinics and groups.

The first survey was from the Medical Computer Services Project of the Washington/Alaska Regional Medical Program in Seattle. It reported on hospitals using or planning to use the computer in Washington and Alaska. These are their findings:

TABLE I

<u>Medical Computer Service</u>	<u>In Operation</u>	<u>Planned</u>
Accounting/Administrative Application	43	9
Medical Records/Information Retrieval	10	17
Laboratory Data Processing	6	10
Food Management/Dietetics	2	3
Computer Aided ECG Analysis	0	9
Automated Patient-History-Taking	1	6
Multiphasic Screening	0	7
Computer-Aided Patient Monitoring	0	6
Computer-Aided Diagnosis	0	4

A second more brief survey was from "Medical Group News". The editor, Milt Golin, polled 722 medical groups across the nation. The results indicated that the computer was involved as follows:

TABLE II

Accounting/Administrative Applications	26%
Laboratory Data Processing	15%
On-Premises EDP for Accounting	11%
Medical Records/Information	2%

Analysts from Manley Management & Marketing Services³ of Greenwich, Conn. predict that the medical market for EDP hardware, peripheral and software products will develop and pass the \$1 billion mark by the end of the Seventies. Those at Stanford Research Institute in Palo Alto, Calif. look for an \$850 million medical EDP market by 1980.

Another Palo Alto firm, Creative Strategies, Inc.⁴ foresees that in the four major market segments of computer applications in medicine: laboratory data systems, electrocardiograph analysis systems, automated patient history systems and hospital information systems, there will be an annual growth rate of

80% over the next five years as a five-year period of R & D begins to pay off.

They predict the installed value of laboratory data systems will grow from a 1970 total of \$12 million to over \$160 million in 1975 and that patient history systems will hit an \$80 million market volume in 1975.

Do my observations as Editor of "Computer Medicine" and observer of the medical scene bear out the predictions of these and other analysts? Is Computer Medicine really in a "Phase II" and diffusing out through society? Are the market researchers too optimistic? What do I see in the future?

1. Do I agree with the predictions of the analysts?

Here, let me make a few qualifying comments. I am not a market research expert. I am primarily a clinician - a specialist in family practice. A genuine family doctor with some unique technical training as an engineer and a medical writer who has had some special experiences in the computer and pharmaceutical industries.

I have had, however, firsthand experiences with market research technics both while I was Medical Director of Dorsey Laboratories, a pharmaceutical division of Sandoz/Wander and as Vice President of COMMED, a subsidiary of COMRESS, a major computer software company.

My observations are made from a very simple technic I have devised as Editor of "Computer Medicine". I call it a "hit". Items come to my desk from a variety of sources: newspaper clipping services, P.R. releases from industry, medical and computer journals, private letters, medical meetings, telephone calls, etc.

If the item describes a bona fide medical computer application that is actually happening, already completed or sufficiently real to be described in a contract, grant or report I call it a "hit".

The categories of hits are seen in Table III. If you will pardon me, I call it my "Hit Parade".

TABLE III

<u>Type of Computer Application (1972)</u>	<u>Jan.</u>	<u>Feb.</u>	<u>March</u>	<u>Total</u>
ECG Analysis System	11	13	8	32
Laboratory Data System	14	7	3	24
Hospital Information System	17	4	2	23
Accounting/Administrative	11	4	0	15
Medical Records/Information	7	8	4	19
X-Ray Diagnostics and Nuclear Medicine	11	7	2	20
Monitoring & ICU Systems	7	6	2	15
Computer-Assisted Dx	4	4	1	9
Automated Patient Histories	2	2	2	6

TABLE III (cont.)

<u>Type of Computer Application (1972)</u>	<u>Jan.</u>	<u>Feb.</u>	<u>March</u>	<u>Total</u>
Multitesting (Multiphasic Screening)	2	4	2	8
Biomedical Data Analysis	6	1	3	10
Computer-Assisted P.E.	1	0	0	1
Medical Network System	2	1	0	3

My results, although different in scope and intent than the Washington/Alaska R.M.P. and "Medical Group News" surveys, are roughly parallel to them. There are several exceptions:

1. Much current activity in automated ECG services put it at the top of my list while the Washington/Alaska survey had it fifth.
2. I do not see as much current activity in automated medical histories in the Seattle report or the analysis of Creative Strategies of Palo Alto.
3. I see new activity in X-ray diagnostics, nuclear medicine and multitesting (multiphasic screening) that were not on the scene a year ago.

2. Is Computer Medicine in Phase II and diffusing into our society?

When I see "hits" cross my desk with datelines like Baraboo, Wisc., Broken Bow, Nebr. and Jacksonville, N. C., I know diffusion is taking place. Baraboo now has an automated ECG service at St. Clare Hospital, Broken Bow has its computerized rural health project and Jacksonville has a prescription service run by a computer. And, I might add, the city where I practice, Herndon, Va., is not exactly on the list of America's medical centers and yet we use automated ECGs, automated patient histories and automated lab facilities! These towns and some of the cities involved in recent medical computer applications are not limited to the Megapolis.

To me there is ample evidence that what was "going to happen" in the decade of the Sixties is now happening. The Innovators set the stage and now the Implementors are hard at work in the decade of the Seventies.

3. Are the market researchers too optimistic?

Although I described earlier in this paper that the time-frame between idea and implementation is getting shorter and shorter in this Age of Technology, I believe that it will take much longer to develop the medical markets than is usually predicted.

One reason was well stated recently by Dr. G. Octo Barnett⁵ when he addressed the National Academy of Engineering in Washington: "One must be careful, however, not to focus too strongly on technology in a mechanical sense, as a unique solution to the problem of increasing medical productivity. The medical industry has rightly been categorized as the last of the

major push-cart industries. A technological innovation which did not include consideration of significant changes in management and organization would be equivalent to adding ball bearings and rubber tires to the push cart, but maintaining the same pattern of buying and selling."

A second reason is that the economics of the health industry is so different from other industries. Hospitals and other non-profit health care units have a constant constraint on the amount of capital funds available for use. Private industry can, as corporations, sell equity shares. Hospitals must usually solicit donations to raise equity capital. Thus, depending on the ability of the hospital to attract donations, its capital may cost more than that of a corporation raising funds in the capital market.

Another reason for a relatively slow inroad for automation is that during the Sixties the results obtained from Hospital Information Systems were generally disappointing. The disappointments were due to such things as liability and high start-up costs of computer systems, the awkwardness of programming languages then available, the lack of enlightened and high quality EDP management in low paying hospital jobs and the reluctance of the professional staff to embrace unfamiliar procedures.

4. What do I see in the future?

During the First National Conference on Computers in Medicine held in Chicago four months ago, I directed a question to Dr. E. R. Gabrieli, director for the Society for Computer Medicine and pioneer in health record networking at the State University of New York in Buffalo. I asked what was the most significant thing he saw or heard at our Conference. He said, "The students at this meeting!"

Medical, nursing and information science students from several Chicago schools were much in evidence and Gabrieli went on to add that for the first time students are now coming into medical school with backgrounds in computer science, programming and various other EDP experiences. Gabrieli commented, "These students will insist on using the computer when they are in practice. They won't have to be talked into it."

I contend, as do others in the field of computer medicine, that when those students, currently in professional schools, graduate they will produce the big change in medical automation.

The physician and nurse of the future will have to efficiently handle many individual patients and large amounts of data. As medical care will come to depend less on the individual physician and more on the coordinated activity of large numbers of professional and auxiliary helpers, timely communication will be crucial. A growing interest in preventive medicine and the management of chronic disease will make long term storage and retrieval of health data a critical need. This coupled with the greater mobility of patients will make it essential that records be complete, easily read and rapidly transferred. The present handwritten disorganized narrative scrapbook now called a health record might even be considered illegal by then.

In the meantime, though, as those students are finishing school, those of us currently involved in Computer Medicine: Phase II have several specific tasks. One with a high

priority is to reach a decision soon on a language for computer medicine. SNOMed, a language patterned after SNOP (Systemized Nomenclature of Pathology) looks like the best bet. An important meeting regarding this was held in Phoenix, Ariz. in January, 1972. Progress is being made toward acceptance of this user language by professional groups such as the American Medical Association, American Hospital Association, American Medical Record Association, etc.

A high priority tag must be attached to realistic plans for funding R & D in medical computer applications. Some industries devote 3 to 8 % of gross sales to R & D. Yet, with an annual expenditure of \$70 billion for health care, I am told that the National Center for Health Research and Development has a budget of only \$70 million or 0.1% of gross expenditures. Hospitals get most of their revenue from insurance companies, and I am not aware of any who reimburse hospitals for R & D in health care. Hospitals thus have little incentive to do R & D to improve productivity. Efforts at changes will have to be made in that sector by all of us in this audience.

And, finally, priority must be established for a wide range of educational activities that will bring students, physicians, nurses, hospital administrators and other health care professionals the necessary understanding and skills to use the computer in creative and constructive ways. Ways that will help improve the medical resources of our nation.

In conclusion, I see much evidence that Computer Medicine has gone into a new era, one that I call Phase II. It is identified by activity and applications away from the established research and medical centers. It is identified by such an increase in the numbers of people interested in the field that the Society for Computer Medicine has been established and a publication, "Computer Medicine", is published monthly. Both are intent on helping health care professionals determine that technology must serve rather than regiment health services.

A few months ago, an acquaintance of mine, a physician who is working full time in the new specialty of computer medicine, went back to his old home town and ended up visiting his old high school. It looked the same, even smelled the same and who should he see but his old English teacher. She looked the same, too, even though 20 years had passed. When she was told that my friend, the doctor, was in computer work, the teacher asked, "Sales or service?" He stood stunned for a minute and then brightly replied, "Service, I hope."

The Innovators of Phase I set the stage for today. They dreamed dreams of high goals and service to mankind. Now, as the Implementors of Computer Medicine: Phase II enter the scene, the dreams of service are becoming reality. In closing, may I remind you that, "Only Service Leads to Greatness".

1. O. G. Barnett, New Physician, 19(2), 119 (Feb. 1970).
2. A. Toffler, Future Shock, p. 27 (Bantom Books, Inc., New York, 1971).
3. Data Management, 9(12), 14 (Dec. 1971).
4. Ibid.
5. O. G. Barnett, Address, Seventh Autumn Meeting, National Academy of Engineering, Washington, D. C., Nov. 1-2, 1971.

PROGRAMMING SYSTEMS FOR OPTIMAL RADIOTHERAPY:
PROGRESS AND PROSPECTS

Judith M. S. Prewitt
Division of Computer Research and Technology
National Institutes of Health
Bethesda, Maryland, U. S. A.

Presented by invitation at The Second Oak Ridge Symposium on
Sharing of Computer Programs and Technology in Nuclear Medicine
April 22, 1972
Oak Ridge National Laboratory, Oak Ridge, Tennessee.

Abstract

Radiotherapy is a problem area in medical decision-making and health care delivery that has already benefited from computer-based interdisciplinary efforts. Significant advances in speed, cost and precision of administered therapy have resulted from successfully automating dosimetry. However, development of new rational bases for optimal radiotherapy requires decision theory and modeling applied to cytokinetics and radioresponse in normal and malignant tissues. Simulation, supported by a conversational, modular, extensible programming language, offers enormous potential for studying clinical implications of proposed treatment regimens. With such a system, the therapist himself can put validated theory into actual practice, and make the computer truly an adjunct to day-to-day medicine.

PROGRAMMING SYSTEMS FOR OPTIMAL RADIOTHERAPY:

PROGRESS AND PROSPECTS

Judith M. S. Prewitt

Despite the strides of the past two decades in both knowledge of the nature of malignant tumors and knowledge of biological response to irradiation, there has been no effective coordination and integration of these advances within a clinically-oriented framework. As a result, there has been no parallel breakthrough in the efficacy of radiologic treatment of these diseases. The most significant recent achievements in treatment are due more to early detection and diagnosis than to better radiotherapeutic technique per se. Control of cancer of the uterine cervix, for example, has improved largely because of the success of mass semi-annual Pap smear screening campaigns. In general, the survival rates in many kinds of tumors which are candidates for radiotherapy not only are distressingly low, but have remained virtually unchanged for many years. The need for new and better approaches to radiotherapy of cancers of grave prognosis is manifest (Silverberg and Holleb, 1971, 1972).

Beyond experimentation with untried types of radiation such as very high energy particles (Hammond, 1972) and combined radio- and chemotherapy (Mitchell, 1968), one possible and promising avenue of approach involves in-depth, extensive application of computer modeling and mathematical optimization to fundamental problems in cancer research. Rather than advocate automation of admittedly inadequate extant therapeutic maxims, we emphasize the implementation of a flexible programming system for optimizing radiotherapy which can bridge the information/communication gap between contemporary radiation research and current radiologic practice. The programming system we envision would provide a medium for (1) simulation and modeling to supplement experimentation and to explore radiotherapeutic hypotheses, and (2) optimization of treatment protocol and treatment delivery. It would nevertheless be a habitable, modular programming system, giving the radiotherapist direct and convenient access to the power of modern computers and auxiliary devices. In short, it would allow the radiotherapist himself to put validated theory into actual practice.

The use of computers in radiotherapy is no innovation, and dates back to 1955 (Tsien, 1955). This early acceptance and appreciation of digital technology was in response to pressing needs of collaborating radiophysicists for rapid, accurate, automated calculation to replace the tedious, time-consuming, error-prone manual methods for preparing individualized treatment plans. The digital computer was indeed constructive in improving technique and precision, while simultaneously curtailing cost and planning time.

By 1966, the role of computers in radiotherapy had been firmly justified, mainly because of contributions to dosimetry and treatment plan visualization. Programs for teletherapy and brachytherapy, written principally in Algol and Fortran, were available for every suitably large European and American computer (I.A.E.A., 1966 and 1968). Economic as well as technological factors were no longer drawbacks. For example, a moving external field dose plan (equivalent to the resultant of a 20 fixed-field combination calculated for 300 points) could be computed on the SAAB D-21 electronic computer in 5 minutes at a cost of approximately \$20, whereas manual determination took 900 minutes and cost approximately \$30. Data preparation, drawing of isodoses, and nonprofessional services added about \$1 in both cases (Ragnhult, 1966). A more careful study might also consider program inefficiency, computer installation idiosyncracies, unjustified computational precision, etc., and favor automation even more.

Facility in dose calculation is prerequisite for any serious efforts toward optimal radiotherapy. The utility of complete dose distributions as compared to sparse field samples is clearly indicated. High correlation between tumor recurrence and low local dose, and between necrosis and high local dose has been verified (Fletcher and Stovall, 1962). In view of characteristically high and nonlinear gradients of tissue response, accurate delivery of dose is critical. For example, a dose of 220 rads to a typical tumor consisting of 10^{10} cells renders an 80% probability of tumor sterilization; a 10% reduction in dose reduces the probability of cure to 26%. Only with the aid of electronic computers can the radiotherapist obtain a complete and accurate dose distribution for each patient at an acceptable cost-benefit ratio.

The headstart in automation has given optimization in radiotherapy at best a modest advantage over similar endeavors in other areas of health care delivery. Computer applications now do encompass several activities relevant to the ultimate goal of providing complete, automatic service and support. These include:

1. external beam radiotherapy (teletherapy)
2. interstitial and intracavity implantation (brachytherapy)
3. visual display of dose distributions
4. record keeping
5. data analysis: retrospective and prospective
6. automation of treatment and auxiliary equipment
7. system analysis of the overall functioning of a radiotherapy center.

The type and extent of computer access required in clinical radiotherapy obviously will vary with the scope of service envisioned, economic realities, and local attitudes toward automation. Both on-site systems (Cox et al, 1967), linked as necessary to a large computer (Haybittle, 1968), and remote treatment planning services (Siler et al, 1964) are commercially available.

Yet the impact of computers on monitoring, controlling, and optimizing radiotherapy has fallen short of expectations and needs. This is certainly not due to a dearth of computer accessibility, both direct and remote, in radiology treatment centers, and likewise not to a deficiency of the necessary mathematical expertise. Rather the scantness of progress stems from substantive considerations relating to foundations.

In particular, the radiotherapeutic maxims or procedural guides now followed in clinical practice are motivated by intuition and supported by relatively crude radiobiological knowledge (Emery, 1966; Perry, 1966). The accepted rule, "Uniformly and maximally dose the tumor to normal tissue tolerance" (Paterson, 1963), has not been restated and formalized as a concept of treatment optimality based on the differential radioresponse of healthy tissues and neoplasms. Neither has it been critically evaluated and validated in even its present preformal statement.

This state of affairs has fostered the development of heuristic techniques supported by computer graphics (e.g. Holmes, 1970; Cox, 1967) for interactively "tuning" treatment plans to un verbalized criteria. "Visual" optimization allows the radiotherapist to continually vary parameters of the treatment plan, immediately observe their effects on the dose distribution, and by repetition, ultimately select an acceptable equipment configuration for therapy. Observer participation in computer-assisted decision-making under uncertainty and risk is unquestionably useful when criteria for success are hard to formalize, but easy to illustrate. But undue fascination with graphical aids can perpetuate reliance on obsolete notions of optimization, and distract us from developing a rationale for radiotherapy based on the growth kinetics and dose response of irradiated tissues. Obviously, the radiotherapist, working manually or even interactively, can review merely some of the relevant plans. Only with formal computer-automated optimization can he compare the merits of all treatment configurations, consistent with radiophysical and radiobiological constraints.

We believe that unique and valuable computer contributions can currently be realized in two important areas: (1) strategy:- development and adoption of a formal, radiobiologically meaningful concept of therapeutic intent, and (2) tactics:- optimizing individually-tailored four-dimensional (space plus time) treatment plans. The basis for such an approach is to exploit fully the fundamental radiobiological phenomena, different in normal tissue and tumors, which make radiotherapy succeed. Although the greater the dose, the greater is the probability of eradicating the tumor, an upper limit is placed on the magnitude of administered dose by the radio-tolerance of healthy tissue in the irradiated volume. Since the therapist cannot preserve the structural and functional integrity

of healthy tissue completely, he must compromise and settle for a trade-off between maximizing the likelihood of cure and minimizing the probability of undesirable and unavoidable complications of radiotherapy. In doing this, he can manipulate the spatial and temporal distribution of dose to great advantage. The treatment planning problem thus resembles a contest or game with nature, and calls for an optimal strategy based on decision theory.

The effort we envision calls for multidisciplinary contributions. In the computer and mathematics domains, these include techniques for data presentation, representation, analysis, inference, and interpretation to support decision-making, as follows:

1. a programming language and executive system that provides a medium for smoothly implementing the entire effort and is (i) interactive, (ii) conversational, (iii) modular, and (iv) extensible;
2. dosimetry algorithms and geometric transformations;
3. representations for two- and three-dimensional dose distributions and anatomical relationships;
4. computer graphics for two- and three-dimensional display of dose and anatomy;
5. computer simulation of growth kinetics and radiation response for normal tissues and tumors (growth-dose-survival models);
6. radiation absorption and scatter models, including inhomogeneity corrections in two- and three-dimensions;
7. image processing and spatial reconstruction techniques for tumor detection and localization;
8. long- and short-term survival studies, involving careful and complete record keeping and statistical analyses;
9. assessment of the degree of precision required for dosimetry (in view of instrumentation errors, mechanical uncertainties in positioning patients, and equally important, in view of dramatic rates of change in response with respect to changes in dose levels, energy, fractionation interval and oxygen levels);
10. multi-stage decision theory, applied to an objective function or figure of treatment merit; and
11. mathematical programming for linearly constrained nonlinear optimization.

A broad plan for optimal radiotherapy would be impossible without a strong, supportive programming environment. A language having the properties enumerated above is not merely an academic curiosity. From the standpoint of applications, the entire system must be characterized by ease of use, ease of change, and ease of growth.

Interactivity and conversationality permit the therapist himself to develop and utilize a treatment planning system in a natural, unencumbered way. Modularity ensures that basic programs need not be replicated. Moreover, deletion and addition of programs and alteration of mathematical models can be made with no more involvement in programming details than is necessary, and with little disruption of other system components.

Program modules can be written with all the precision and efficiency that is appropriate and feasible, but they must be interlinkable via the executive program.

Extensibility allows for conceptual growth and new capabilities, as well as augmentation of the data base. A priori knowledge about cytokinetics, tissue growth and radiobiology can be incorporated in the data base and models as it accrues. Treatment-relevant concepts can be defined and explored on-line. Radiophysical and radiobiological constraints and optimality criteria expressing treatment sub-goals and end-goals can be modified with experience and clinical evaluation. Thus, the whole treatment planning system remains responsive to contemporary radiotherapeutic rationales without excessive reprogramming.

This kind of programming system obviously incurs a substantial overhead, but it materially improves human utilization of information resources and human efficiency. It is likely also to apply to other treatment planning problems, and therefore seems well worth the effort.

Although we hardly believe that this type of effort -via simulation and modeling, image processing, and optimization- is the only possible approach, we feel that it is efficient, general, and reasonable. It offers enormous potential for systematically and economically evaluating a broad range of treatment variables, strategies, and tactics in the context of plausible radiobiological mechanisms.

A full discussion of the role of mathematical models in radiation treatment planning is reserved for a future paper. For present purposes, it suffices to say that models will usually consist of differential or integral equations, and be either the expected value type or the stochastic type. They will be simulated by algorithms which must be initialized by externally supplied radiobiological and radiophysical parameters.

Models serve a broader purpose than just a posteriori simulation. They can be as useful for prospective studies as for retrospective analyses (where they are mainly explanatory). In this new capacity, they can suggest definitive experiments for choosing among competitors based on sensitivity analysis of model parameters. They can also suggest optimal experimental designs (data sampling plans) for estimating the models parameters with, say, minimum variance or maximum utilization of the total experimental effort. Finally, they can be used to ascertain the precision and accuracy requirements on calculated dose and delivered dose.

By treatment planning in cancer radiotherapy we mean systematically and rationally designing a regimen of deliberate and controlled radiotherapeutic intervention in the course of disease. Radiotherapeutic strategy is based on the differential response of normal and malignant cells and tissues to irradiation. The radiotherapeutic ideal is to completely eradicate the tumor while simultaneously sparing healthy tissue. Radiotherapeutic tactics involve generating and shaping an appropriate time

sequence of patterns of absorbed dose in the tumor-bearing volume. Shaping is the result of additively combining the intensities of spatio-temporally coincident radiation.

The clinician tries to control the spatial distribution of delivered radiation so that the effective portion of the resultant dose pattern conforms to the target as well as possible and avoids vulnerable non-target regions. He tries to control the temporal distribution of delivered radiation so that the differential radioresponse and recovery characteristics of normal and malignant cells during repeated or fractionated exposure to radiation is exploited. The treatment parameters at his disposal include: (1) the kind and number of external beams and internal implants with possibly different field dimensions, intensities, energies, and gradients; (2) the configuration of the selected radiation sources in relation to the patient contour, the vital organs, and the tumor; (3) relative times and duration of exposure or application; (4) various field modifiers; (5) instantaneous dose rate; (6) radiation quality; and (7) auxiliary chemotherapy.

Radiotherapeutic technique has evolved empirically over the past 70 years. Current methods may be efficient, but they are not necessarily optimal. They are succinctly summarized by the simple treatment maxim: "Uniformly and maximally dose the tumor to normal tissue tolerance"(Paterson, 1963). Treatment parameters are manipulated accordingly.

In general, it is true that the greater the delivered dose, the greater is the likelihood of eliminating clonogenic tumor cells. However, the magnitude of the applied dose is limited above by the radiation tolerance of the normal tissue in the irradiated volume, and limited below by the radioresistance of the tumor. The radiotherapist therefore cannot succeed completely in preserving healthy tissue, and he must settle for a trade-off between maximizing the likelihood of cure and minimizing the likelihood of treatment morbidity and mortality.

The probability of curing the tumor is strongly dependent on cell vulnerability, cell repair rate, and tissue repopulation potential as a function of many complex inherent predispositions and environmental conditions (Tubiana, 1971). These include the obvious factors of local cell density, local radiation dose, and between-exposure interval, as well as subtle factors such as the ratio of quiescent to proliferating and non-viable cells, and their spatial distribution; the distribution of the proliferating fraction in the various phases of the mitotic cell cycle at the time of irradiation; metabolic conditions; and the bio-chemical milieu. Because of the functional forms of these dependencies, small variations in the local radiation dose and timing have a disproportionately large influence on the probability of cure. This natural restriction on effectiveness of delivered radiation underscores the desirability of determining the most effective and acceptable spatial and temporal distribution of dose as precisely as possible, and adopting this pattern as the treatment policy.

The previously quoted treatment maxims do not adequately explicate the subtleties of the decision-theoretic point of view, as applied to radiotherapy. In pursuing this approach properly,

the clinician must deal with probable dose responses rather than absolute lethal dose levels. We therefore revise the radiotherapeutic ideal as follows: optimal treatment achieves the highest probability of eradicating the tumor compatible with the highest probability of retaining structural and functional integrity in normal tissue.

From the preceding discussion, it is clear that construction of individualized optimal plans requires treatment-specific, patient-specific and general information, formatted for convenient entry yet efficient computation. Treatment-specific information includes: (1) dose distributions as functions of source-to-skin distance, treatment machine geometry, and energy for the contemplated radiation sources; (2) beam positioning constraints, if any, and (3) characteristics of modifying filters (wedge angles and absorption coefficients); as well as (4) corresponding tissue radiosensitivity data, absorption coefficients and scatter ratios. Patient-specific information includes: (5) anatomical contours and (6) tissue composition and cell density for the body cross-section, the tumor, and vulnerable vital regions. The decision-theoretic approach also requires general information consisting of: (7) biophysical models (growth kinetics and dose-survival) and constraints for irradiated malignant and normal tissues, from which the probabilities of treatment outcomes can be computed; (8) utilities and penalties ascribed to the desirable and deleterious outcomes; and (9) directives indicating which optimality principles (local and global) are to be invoked.

The fundamental problem in radiation treatment planning is dosimetry: determining local dose in an absorbing and scattering medium. Single field data are necessary for the calculation of composite distributions. Mathematical methods have been used extensively to minimize the number of experimental values needed and to obtain generality.

In teletherapy, patient contours are usually expressed in a polar coordinate system originating at the tumor "center" (Tsien, 1955; Jones, 1956). For convergent beam therapy (both fixed and rotational), this origin is also the isocenter or center of rotation. For nonconvergent or multifocal beam configurations (a technique especially useful in shaping nonconvex "butterfly" isodose patterns that conform to tumor but circumvent adjacent vital organs), the origin is symmetrically placed relative to the foci (Sterling and Perry, 1964). The grid generally consists of a polar lattice with rays at 5 or 10 degree intervals, and radial increments of 1 centimeter or 1/2 inch with an overall radial span of 40 centimeters. This mesh is fine enough to meet current error tolerances for dosimetry, and is also consistent with available high-speed computer storage. Radial values need not be equi-angularly spaced; in our own system (Prewitt, 1972), missing values are interpolated.

Dose distributions are also ultimately referred to this primary coordinate system. The total (single-exposure) dose at any point in the treated region consists of a linear combination of the individual beam contributions, weighted by the appropriate relative intensities. Each component distribution, however, is specified in a more favorable secondary polar coordinate system

which is standard for all beams of the same type and operating characteristics (e.g. same source-to-skin distance). Each single-field distribution must be translated and rotated prior to combination with other constituents, but after correction for modifying filters (wedges), oblique incidence (entry angle and body curvature), and tissue heterogeneity. Similarly, organ and tumor outlines are entered in convenient secondary polar coordinate systems, to which the composite dose distribution, described in the primary coordinate system, can be transformed as needed.

Analytical difficulties are encountered when trying to maintain accuracy in geometrically transforming contours, isodoses, and entire dose distributions from one polar coordinate system to another. This occurs because the Jacobian of the coordinate transformation is position-dependent, and has prompted some investigators to use Cartesian coordinates in place of the easier-to-visualize, treatment-oriented polar system. The Cartesian grids are also better suited for display on lineprinters.

Individual beam fields are either taken from dose atlases, precomputed, or calculated on-line. Precomputed distributions are digitized and entered via keyboard or card reader (punched card or magnetic card), or else recalled from tape or disk storage. Three data formats have been used for representing beam fields: (1) isodose curves, (2) transverse beam dose plots, and (3) decrement lines. The characteristic shape of any of these representations is determined by the energy of the incident radiation, the field area, and the geometry of the collimating system. For some purposes, source-to-skin and source-to-wedge distances, entry angle, and portal dimensions (or angular field width), must also be specified explicitly. Preferences are dictated by the usual storage-computation trade-offs.

Isodose curves consist of a set of isopleths for field intensity values in a plane through the beam central ray. These isopleths are superimposed on a polar grid with origin at the source. Dose values should be most densely sampled off-axis toward the periphery of the field, where spatial rates of change are greatest and least uniform.

Transverse beam plots give the dose distribution at a fixed depth orthogonal to the central axis of an isodose chart. Dose is given as a different function of off-axis position for each depth level.

The decrement line system is based on the existence of a family of surfaces coaxial with the center ray, each of which is the locus of points at which dose is a constant fraction or decrement of the central axis depth dose at the same depth. For the radiation utilized in therapy, it has been determined experimentally that a plane through the central axis intersects each isodecrement surface in paired straight lines: the decrement lines. To specify a decrement line, only three numbers are required: the percentage decrement, and the positions of just two points on that line. The representation is obviously very economical.

A great diversity of dosimetry algorithms exists for external beam therapy. All of the algorithms partition the dose at any point in the absorbing medium into the sum of contributions due to the primary beam and to scattered radiation (Johns, 1951). From the preceding discussion of representations, it is clear that the dose value at any point in a uniform medium can be derived from the product of two functions: the first representing the central axis depth dose, and the second representing the transverse dose distribution. The central axis depth dose (for beam energies exceeding 1 MeV) varies logarithmically with depth and can be expressed by an exponential function.

The variety lies in the representation of the transverse dose function. The methods are of three principal types: (1) empirical curve fits (using mathematical expressions with no apparent physical basis), (2) experimental or semi-analytical curve fits (using some analysis of experimental data from a physical standpoint), and (3) theoretical or analytical formulas (using first principles to derive an expression for dose in an absorbing and scattering medium).

The empirical curve fit (Sterling, 1964) adopts a convenient formula, the cumulative normal probability integral, to represent transverse dose as a function of field dimensions, source-to-entry point distance, and a treatment machine geometry factor. Algorithms in the second category use interpolation based on cross plot data for two depths (Van de Geijn, 1965), or what is equivalent, decrement lines (Orchard, 1964; Orr et al, 1964). Off-center corrections can be applied to tissue-air ratios in place of central axis depth dose (Siler et al, 1965; Holt et al, 1966). Algorithms in the third category spatially integrate scattered radiation contributions to each point (Clarkson, 1941; Meredith and Neary, 1944) for any shape field, utilizing scatter-air ratios derived from tissue-air ratios (Gupta and Cunningham, 1966). As an alternate to direct spatial integration, the dose at a point may be expressed as the product of two exponential functions: the first representing central axis depth dose as before, and the second representing the orthogonal or transverse dose distribution, again as the resultant of primary and scattered radiation (Schoknecht, 1966).

Inaccuracies reported for these algorithms range from about ± 1 or 2% at tumor center to as much as ± 4 or 5% at the beam penumbra, and are somewhat higher when wedge filters are included in the calculations (IAEA, 1966). An error of approximately 5% in the tumor maximum incurred with these methods can be accompanied by a 20 to 25% uncertainty in an intended dose of 20% tumor maximum to a vital organ. The neglect of tissue heterogeneities can foster errors of 20 to 30% for calculated dose based on absorption and scatter in homogeneous water phantoms. This suggests that confidence intervals for treatment plans might be helpful.

These dosimetry algorithms differ in their facility for handling body curvature, tissue heterogeneity, contour irregularity, compensating filters, and three-dimensional

extensions. In selecting one method rather than any other, we appeal to the following criteria:

1. accuracy
2. suitability for (digital) computation
3. speed of computation
4. versatility and generality
5. portability (machine independence)
6. suitability for inclusion in:
  - a. interactive systems for planning acceptable treatment
  - b. automated systems for planning optimal treatment.

These are general qualities of algorithms that we may desire for any evolutionary scientific endeavor in which final solutions are still to come. Cunningham's algorithm appears to be the method of choice.

As a minimum, final results consist of the dose distribution. Sometimes the coordinate grid, patient outline, organ contours and tumor are superimposed, and positions of radiation sources are indicated. Data is presented on either a lineprinter, teletypewriter, incremental plotter, or CRT graphic display. Printout of the matrix of total doses can be numeric, symbolic (comparable to digitized isodose plots), or pictorial (with simulated graytones representing field intensity). The incremental plotter generates hard-copy continuous isodose curves and anatomical contours, with identifying legends. The CRT produces the same as soft-copy. It is difficult to obtain customary true-to-scale dose distributions on the lineprinter. Future systems will require output of the entire treatment sequence.

Perhaps the most difficult yet most crucial aspects of automating optimal treatment planning and delivery are the following:

1. the articulation of criteria for judging therapeutic value;
  2. the subsequent transformation of these verbal, preformal criteria into quantitative, formal figures of merit; and
  3. the enumeration of relevant treatment constraints.
- The cost-benefit-effectiveness ratios of automated treatment planning depend heavily on the adequacy of the figure of merit for explicating therapeutic worth, and on the computational complexity which the selected measure entails.

Some of the simpler clinical procedural maxims have already been adopted as the basis for computer-automated treatment planning in the operations-research style score function approach of the British (Hope et al, 1966, 1967, 1970), and in the more analytic linear programming approach of the Americans (Bahr et al, 1968). Both of these attempts are important, but only partial contributions. They focus on the signs and symptoms of good and bad plans rather than on the causes underlying therapeutic success and failure. Since the treatment maxims behind them do not necessarily reflect radiobiologically optimal strategy, neither do the computer implementations. Much of value, however, can be learned from their strengths and weaknesses.

The score function is a suggested measure for evaluating a treatment plan in terms of six plausible empirical criteria:

1. dose gradient across the tumor (uniformity of dose)
2. dose to tumor relative to maximum incident dose (treatment efficiency)
3. integral (total) dose
4. shape conformity of treated area relative to chosen treatment area
5. dose to vulnerable or radiosensitive healthy regions
6. dose to regions of possible direct extension or lymphatic spread.

These six attributes or features are quantitated and combined nonlinearly in a polynomial which provides a single summary score or purported figure of merit for an individual treatment plan.

The score components:

1. dose gradient score
2. tumor dose score
3. integral dose score
4. shape score
5. vulnerable dose score
6. entry point score

are actually complicated functions of the basic controllable treatment variables and largely uncontrollable radiobiological parameters.

The score function has been used to manually review and compare suggested treatment plans. In principle, it could be incorporated in a computer-based optimization scheme. For the present analytical form of the score function, this could be accomplished by adaptive learning techniques of pattern recognition. Polynomial coefficients would be iteratively adjusted, until the best match is obtained between the rank order of score function values and the rank order of therapists' judgments on the same training set of treatment plans. The score function with these coefficient values could then be applied to any other contemplated treatment plan. There is no guarantee, however, that the function will reproduce therapists' judgments as satisfactorily in the future as it was "tuned" to do in the past. Moreover, there is no guarantee that all the important radiobiological variables and intuitive cues are adequately, if at all, taken into account.

Although the score function can be faulted on nontrivial grounds, it is historically important as the first systematic attempt to quantitate desirable attributes of radiation treatment plans, and to incorporate them in an objective function suitable for numerical optimization. Its deficiencies, however, should not go uncriticized, and are actually instructive.

First, there is no explicit connection between score function criteria and radiobiological phenomena at the cell and tissue level. Whereas tolerance dose levels and dose conformity to target, simultaneous with avoidance of vulnerable regions, are obviously relevant, there is no reason to assume that a uniform dose is optimal in some plausible sense for all tumors. On the contrary, mathematical modeling shows that this treatment criterion is logically inconsistent with certain reasonable cell survival-oriented concepts of optimality. In particular, for tumors in which there are spatial gradients of both cell density

and radiosensitivity, a uniform dose is generally counter-optimal for achieving local tumor control with a prescribed minimum probability of cure (Fischer, 1969). Maximizing the survival rate in irradiated adjacent normal tissue in a manner compatible with sterilization of all or most tumor cells further complicates the analysis.

Second, study of actual score function behavior with variation in treatment delivery variables such as relative beam positions reveals the existence of multiple local optima with different absolute scores. Thus the mathematical properties of the score function preclude unequivocal determination of the unique globally optimal treatment plan. Confronting the radiotherapist with this type of methodological ambiguity or uncertainty is untenable.

The linear programming approach to radiation treatment planning differs methodologically from the score function approach, but invokes similar empirical criteria. The notable exception is substitution of maintenance of stipulated dose ranges for the dose uniformity criterion. However, the criteria are introduced implicitly as part of the linear programming formulation, rather than explicitly. They take the form of geometric constraints and local dosage requirements for tumor and other anatomical regions of interest. The method focuses on spatial optimization of dose for a single treatment. It utilizes an objective function expressed directly in terms of simple controllable treatment parameters (e.g. beam position and relative intensity) rather than mediated by the criteria. In other words, the method pursues an operational definition of optimality.

Consequently, the output of the linear programming method is not a merit score for any proposed plan. Rather, it is that external beam treatment plan among all those considered, which best achieves the proposed objective. This objective is delivery of a (presumably) lethal but not excessive local dose to tumor (tumor dose requirements), with the maximum dose concentrated at the tumor center (also taken as the isocenter), consistent with the following: limited local dose to adjacent tissues (anatomical dose constraints), and linearly constrained minimization of average local dose to vulnerable regions. The tumor lethal dose is defined by lower and upper dosage bounds rather than by a single value. The upper bound corresponds to overkill.

The objective which is minimized is a linear function of non-negative beam weights or relative intensities (corresponding to short-term exposure to radiation). Likewise, the dosage constraints consist of linear inequalities in the beam weights. A finite but large set of beams disposed around the patient is postulated, and linear programming optimally assigns the beam weights. Actual beams correspond to nonzero weights. When most of the weights are zero, the plan calls for multiple fixed beams; optimal beam positions are inherent biproducts of the optimization procedure. When almost all weights are positive, the plan can be interpreted as a call for rotating beam therapy, with possibly positionally varying relative intensities. The

tendency is to closely meet local dosage bounds for tumor and adjacent tissues, but to stay well below in vulnerable regions.

Plausible variations in the objective function and constraints are easily conceived. One such variation is to stress treatment efficiency as measured by the ratio of incident dose to tumor dose, or by the ratio of tumor dose to vital organ dose. Another variation, parametric programming, involves relaxation of the dose constraint for adjacent tissues continuously to the lowest value consistent with obtaining a feasible solution (i.e. at least one positive and no negative weights). This leads to a generally more uniform distribution of dose, but at the price of an elevated dose in areas to be spared and an increased number of entry portals.

Dose distributions obtained by both linear and parametric programming are superior to intuitive "textbook" solutions employing the same constraints, bounds, and objective function, but limited to a specific number of fields with prescribed entires and free weights. Sometimes no feasible "textbook" solution exists; other times constraints are violated; and often spared regions are dosed to tolerance.

Nevertheless, most of the criticisms which apply to the score function approach likewise apply to the method of linear programming. Principally, the emphasis is on optimal implementation of external beam therapy. The simplistic textbook procedural guide of dosing the tumor to healthy tissue tolerance is automated, and radiobiological justification is not broached. (However, on information-theoretic grounds, this is the best strategy to follow when nothing is known in advance about local tumor structure and response.) With respect to implementation, the objective function is actually nonlinear in other important treatment parameters such as wedge angle, source-to-skin distance, entry angle, portal size, and beam energy. Thus more advanced techniques for constrained mathematical programming are required for even simple extensions of the method to more versatile beam arrangements.

The decision-theoretic approach can be instrumental in developing a radiobiological rationale for therapy which averts much of the foregoing critique. For this purpose, we distinguish between radiotherapeutically optimal treatment protocol (strategy) and optimal treatment implementation (tactics) (i.e. executing the optimal protocol as precisely as possible). The first aspect of optimization is concerned with the construction of that treatment plan which best fulfills the radiotherapeutic intent. The results will clearly be conditional upon two factors: (1) current a priori knowledge of radiobiology and cytokinetics, and (2) the selected figure of merit for comparing treatment plans. The second aspect is concerned with determining that equipment configuration from those available which best implements any proposal for tumor management, not merely the distinguished plan generated by the first phase. The results will again be conditional upon two factors: (1) the regions within the patient contour over which the match is to be made and the purported relative importance of these regional matches, and (2) the selected figure of merit for evaluating conformity of the actual plan to the ideal plan.

A suitable figure of merit for design of the optimal plan consists of a linear compound of probabilities for the two principal elemental outcomes of irradiation and their products: (1) the probability of cure (advantage), and (2) the probability of concurrent damage to healthy tissue (adversity). Both probabilities are generally numerically different but similarly S-shaped functions of dose. The figure of merit is a non-linear function of the relevant biological parameters, including critical dose for normal and malignant tissues, differential rates of cell repair and proliferation, and fractionation interval. The compounding coefficients multiplying the probabilities are utilities or penalties.

This objective function has the advantage of including many simple treatment strategies as special cases. It is a generalization of the objective of minimizing the integrated dose to vulnerable regions or maximizing the integrated dose to target regions under anatomical dose constraints. It correctly emphasizes the subjective judgments and competing sub-goals that must be considered systematically in treatment planning along with the objective data. At present, a strict definition of cure is used, namely eradication of all clonogenic tumor cells in the primary lesion. However, the objective function can be weakened if desired by substituting tumor management for tumor cure. This in turn can be expressed by substituting the probability that some tumor cells rather than no tumor cells survive.

A suitable figure of merit for the design of the best equipment configuration consists of a matching coefficient. As an example, this can be a weighted sum of squared differences between the ideal local dose and the local dose realized in the target region and in the non-target vulnerable region as well. For external beam therapy, this figure of merit is non-linear in the treatment variables. These treatment variables include: (1) number of beams, (2) relative source to target location, including angular position, entry point and angle, and source-to-surface distance, (3) type(energy) of radiation, (4) shielding bar geometry, (5) wedge angle, and (6) relative beam intensities.

Decision theory determines the most favorable relationship of cure and damage probabilities (Prewitt, 1972), but not uniquely. The result is actually a one-parameter family of optimal probability pairs. In other words, decision theory determines the optimum up to one degree of freedom, and an additional constraint must be invoked. This can take the form of another external optimality principle or experimentally obtained information. Either constraint introduces radiobiological parameters (e.g. local lethal dose, cell repair rate, cell proliferation rate, cell loss), radiobiological functions (e.g. probable dose response, spatial density of cells), and treatment variables. This new decision-theoretic approach for optimal radiotherapy will be described in a forthcoming paper (Prewitt and Davis, 1972).

With regard to optimally administering radiotherapy, the dosimetry calculations for presently used equipment are well in hand, and can cope with corrections for scattered radiation, tissue inhomogeneities, and three-dimensional extensions. With

regard to optimal therapeutic intervention in the course of disease, although a definitive theory of the radiobiology and kinetics of cell proliferation and tissue growth is not yet available, there is a great deal of unexploited clinical and experimental evidence. This material is coherent enough to motivate construction of cell survival and tissue growth models from which we can ascertain the kind of mathematics needed for optimization. The digital computer required for dosimetry, on-line display, and interactive radiation planning is sufficient for also dealing with mathematical optimization in this broadened conception.

Decision theory is a particularly attractive approach to optimal treatment planning for several reasons. First, it provides a systematic way of relating the multiplicity of tumor growth parameters to controllable treatment parameters. Second, it obliges the therapist to enumerate the possible advantages and adversities of irradiation, and to pass judgment on their relative desirability. Third, it emphasizes the uncertainty and risk associated with treatment results, and the unavoidability of trade-offs among the treatment sub-goals. Fourth, it makes clear that responsibility for establishing preference relations among treatment sub-goals must remain with the therapist.

Summary

High speed digital computers can provide a new medium for evaluating medical decision making and administering health care in a manner that is responsive to state-of-the-art medicine. Computer services in clinical radiotherapy represent a model application of this type which has already benefited from a coordinated interdisciplinary, computer-based effort. To date, these benefits have been in areas which relate primarily to the speed, cost, and precision of administered therapy.

The variety of computers used in radiotherapy is only to be expected. But variety of languages, and especially the use of machine codes, however efficient from the point of view of computation, is inefficient from the point of view of program exchange and sharing.

Further developments will involve modeling biological systems, with special emphasis on mathematical optimization and sequential or multi-stage decision theory. In the near future, we expect to be able to incorporate and utilize radiobiological information. Simulation, set in the framework of an extensible programming language, offers enormous potential for studying the clinical implications of proposed courses of treatment in the light of radiobiological research. Many combinations of key parameters and treatment constraints can provide data on a broad spectrum of representative cases in far less time, and with far less cost, than can field studies and laboratory experimentation. In particular, individual patient responses to changes in treatment variables and physiological state can be charted and can guide the course of treatment.

Progress in utilizing simulation techniques in radiotherapy optimization has been slow primarily because of limited knowledge of the radioresponse of tumors and normal tissue in vivo. As the store of relevant information accrues and is disseminated, development of simulation models will no doubt accelerate. The form of this forthcoming knowledge can at this point be anticipated and planned for. A programming system which provides appropriate mathematical optimization algorithms, along with convenient means for easily changing the a priori data base and treatment optimality criteria, can be an important step toward our goal. It will allow the radiotherapist to construct plausible objective functions subject to realistic constraints. Only in this way can the computer truly become an adjunct to day-to-day medical practice.

Thus having witnessed the invaluable computer contributions to dosimetry and to the design of acceptable treatment protocol with modest computer capacity and programming sophistication, we are confident that a significant practical impact on the design of optimal treatment protocol is also within reach.

REFERENCES

- Bahr, G. K., Kereiakes, J. G., Horwitz, H., Finney, R., Galvin, J. and Goode, K. (1968)
The Method of Linear Programming Applied to Radiation Treatment Planning.
Radiology 91: 686-695, 1968.
- Clarkson, J. R. (1941)
A note on depth doses in fields of irregular shape.
Brit. J. Radiology 14: 265-268, 1941.
- Cohen, M. (1966)
The organization of clinical dosimetry.
Acta Radiologica 4: Therapy Physics Biology, 1966.
I. The four stages of clinical dosimetry. 233-256.
II. Some special topics in treatment planning: 311-335.
- Cohen, M. (1968)
Impact of the Computer on the Organization of Radiotherapy.
In IAEA, 1968: 101-107.
- Cox, J. R., Garth, V. W. and Holmes, W. F. (1967)
The Programmed Console: An Aid to the Radiologist in Treatment Planning.
In First International Conference on The Use of Computers in Therapeutic Radiology, Cambridge, England, 1966.
London, England: Brit. Institute of Radiology, 1967.
- Cox, J.R., Gallagher, T.L., Holmes, W.F. and Powers, W.E. (1967)
Programmed Console: An Aid to Radiation Treatment Planning.
In Proc. 8th IBM Medical Symposium: 179-188, 1967.
- Cunningham, J. R. (1968)
A Generalized Scatter Function for Computer Calculation of Dose in a Phantom.
First International Conference on Use of Computers in Radiation Therapy, 1966.
- Ellis, F. (1968)
Computers and Planning in Cancer Treatment.
In IAEA, 1968: 145-161.
- Emery, E. W. (1966)
Clinical evaluation of treatment plans.
In IAEA, 1966: 177-181.

- Fischer, J. J. (1969)
Theoretical considerations in the optimisation of dose distribution in radiation therapy.
Brit. J. Radiology 42: 925-930, 1969.
- Fletcher, G. H. and Stovall, M. (1962)
A study of the explicit distribution of radiation in interstitial implantations II. Correlations with clinical results in squamous-cell carcinomas of the anterior two-third of tongue and floor of mouth.
Radiology 78: 766-782, 1962.
- Gupta, S. K. and Cunningham, J. R. (1966)
Measurement of tissue-air ratios and scatter functions for large field sizes, for cobalt-60 gamma radiation.
Brit. J. Radiology 39: 7-11, 1966.
- Hammond, A. L. (1972)
Cancer Radiation Therapy: Potential for High Energy Particles.
Science 175: 1230-1232, May, 1972.
- Haybittle, J. L. and Houston, A. A. (1968)
The use of a multi-access computer for radiation treatment planning.
Brit. J. Radiology 41: 927-931, 1968.
- Holmes, W. F. (1970)
External Beam Treatment-Planning with the Programmed Console.
Radiology 94: 391-400, 1970.
- Holt, J. G., Ritter, F. W. and Laughlin, J. S. (1966)
In The Use of Computers in Therapeutic Radiology.
Report Int. Conf. Computers in Radiology, Cambridge, England, 14-17 June, 1966.
London, England: British Institute of Radiology, 1967.
- Hope, C. S. (1970)
Computer Optimization of Treatment Planning.
In Medical Computing: Progress and Problems (M. E. Abrams, ed.): 133-145. New York: American Elsevier, 1970.
- Hope, C. S., Laurie, M. J. E. and Orr, J. S. (1966)
The optimization of treatment plans.
In IAEA, 1966: 118-128.
- Hope, C. S., Laurie, M. J., Orr, J. S. and Halnan, K. E. (1967)
Optimization of X-ray Treatment Planning by Computer Judgment.
Phys. Med. Biol. 12: 531-542 (1967).

- International Atomic Energy Agency (1966)
 Computer Calculation of Dose Distributions in Radiotherapy.
 Report of a Panel held in Vienna, 18-22 October, 1965.
 Technical Report Series No. 57
 Vienna, Austria: International Atomic Energy Agency, 1966.
- International Atomic Energy Agency (1968)
 Role of Computers in Radiotherapy.
 Report of a Panel held in Vienna, 10-14 July, 1967.
 Vienna, Austria: International Atomic Energy Agency, 1968.
- Johns, H. E. (1961)
 The Physics of Radiology.
 Springfield, Illinois: Charles C. Thomas, 1961.
- Jones, D. E. A., Gregory, C. and Birchall, I. (1956)
 Dosage distribution in rotational cobalt 60 therapy.
 Brit. J. Radiology 29: 196-201, 1956.
- Massey, J. B. (1968)
 Impact of Computers in Radiotherapy.
 In IAEA, 1968: 113-125.
- Meredith, W. J. and Neary, G. J. (1944)
 The production of isodose curves and the calculation of energy
 absorption from standard depth dose data.
 Brit. J. Radiology 17: I. 75-82, II. 126-130, 1944.
- Mitchell, J. S. (1968)
 Applied science and the development of radiotherapy-past,
 present and future.
 Brit. J. Radiology 41: 729-748, 1968.
- Orchard, P. G. (1964)
 Decrement lines: a new presentation of data in cobalt 60 beam
 dosimetry.
 Brit. J. Radiology 37: 756-763, 1964.
- Orr, J. S., Laurie, J. and Wakerley, S. (1964)
 A study of 4 Mev transverse data and associated methods
 of constructing isodose curves.
 Physics Med. Biol. 9: 505-515, 1964.
- Paterson, R. (1963)
 The Treatment of Malignant Disease by Radiotherapy.
 Second Edition.
 London, England: Edward Arnold Ltd., 1963.

- Perry, H. (1966)
 Clinical evaluation of treatment plans: Criteria use to select optimum plan.
 In IAEA, 1966: 129-145.
- Prewitt, J. M. S. (1972)
 Nonlinear Analysis In Biomedical Computing for Cancer Research. Journees d'Informatique Medicale (Medical Data Processing Symposium), 28 February-3 March, 1972, Toulouse, France. Rocquencourt, France: IRIA(Institut de Recherche d'Informatique et d'Automatique).
- Prewitt, J. M. S. and Davis, L. A. (1972)
 A decision-theoretic approach to optimizing radiation treatment planning.
 American Association of Physicists in Medicine, 26-29 June, 1972, Philadelphia, Pa.
- Ragnhult, I. (1966)
 Economics of Computer Dosimetry.
 In Computer Calculation of Dose Distributions in Radiotherapy, Technical Report Series No. 57: 146-150. Vienna, Austria: International Atomic Energy Agency, 1966.
- Schoknecht, G. (1966)
 In The Use of Computers in Therapeutic Radiology. Report Int. Conf. Computers in Radiology, Cambridge, England, 14-17 June, 1966.
 London, England: British Institute of Radiology, 1967.
- Siler, W. and Laughlin, J. S. (1964)
 A computer method for radiation treatment planning.
 Comm. Assn. Computing Machinery 5: 407-408, 1964.
- Siler, W., Ritter, F. and Laughlin, J. S. (1964)
 Patient simulation in X-ray therapy.
 Ann. N. Y. Acad. Sci. 115:1038-1048, 1964.
- Silverberg, E. and Holleb, A. I. (1971)
 Cancer Statistics, 1971.
 Ca-A Cancer Journal for Clinicians 21: 13-31, 1971.
- Silverberg, E. and Holleb, A. I. (1972)
 Cancer Statistics, 1972.
 Ca-A Cancer Journal for Clinicians 22: 2-20, 1972.

- Sterling, T. D. and Perry, H. (1965)
 Computation of Radiation Dosages.
 In Computers in Biomedical Research (Vol. 1) (R. W. Stacy and
 D. Waxman, eds.), Chapter 18: 439-464.
 New York: Academic Press, 1965.
- Sterling, T. D. and Perry, H. (1964)
 Planning radiation treatment on the computer.
 Ann. N. Y. Acad. Sci. 115: 976-997, 1964.
- Sterling, T. D., Perry, H. and Katz, L. (1964)
 Derivation of a mathematical expression for the percent depth
 dose surface of cobalt-60 beams and visualization of multiple
 field dose distributions.
 Brit. J. Radiology 37: 544-550, 1964.
- Tsien, K. C. (1955)
 The application of automatic computing machines to radiation
 treatment planning.
 Brit. J. Radiology 28: 432-439, 1955.
- Tubiana, M. (1971)
 The kinetics of tumor cell proliferation and radiotherapy.
 Brit. Journal of Radiology 44: 325-347, 1971.
- Van de Geijn, J. (1965)
 The computation of two and three dimensional dose distribu-
 tions in cobalt-60 teletherapy.
 Brit. J. Radiology 38: 369-377, 1965.

END

RADIATION TRANSPORT CODES FOR POTENTIAL APPLICATIONS RELATED
TO RADIOBIOLOGY AND RADIOTHERAPY USING PROTONS,
NEUTRONS, AND NEGATIVELY CHARGED PIONS^{1,2}

T. W. Armstrong
Oak Ridge National Laboratory
Oak Ridge, Tennessee, 37830

Abstract

In connection with long-term programs in space-vehicle, accelerator, and reactor shielding, the Oak Ridge National Laboratory has developed rather sophisticated calculational methods for determining the transport in matter of protons, neutrons, pions, muons, electrons, and photons. This work has resulted in several Monte Carlo radiation-transport computer codes which are quite general with respect to allowable source-particle descriptions, geometries, material compositions, etc. Recently, a limited study has been directed toward applying some of these computer codes to predict quantities of interest in the fields of radiotherapy and radiobiology, such as depth-dose patterns, and toward assessing the accuracy of the codes for applications in these fields. In this paper, the calculational methods are described and comparisons of calculated and experimental results are presented for dose distributions produced by protons, neutrons, and negatively charged pions. Comparisons of calculated and experimental cell-survival probabilities are also presented.

---

¹Research partially funded by the National Aeronautics and Space Administration, Order H-38280A, under Union Carbide Corporation's contract with the U. S. Atomic Energy Commission.

²Abstract of ORNL-TM-3816, Oak Ridge National Laboratory (in press).

RAPID MANIPULATION AND DISPLAY OF COUNTS VS. TIME CURVES
BY HARDWARE AND SOFTWARE

Thomas F. Budinger

ABSTRACT

Quantitation in nuclear medicine involves fast data acquisition, rapid processing, and display of 4k to 16k data point arrays. Curves of counts vs. time over selected organs or body regions are the output for clinical and research observations. These curves can be generated, manipulated, and displayed conveniently on a small computer system by using hard-wired, system program, or user software. Flexibility and speed for performing any suggested clinical or research time function manipulation, including Fourier transformation, can be accomplished in less than one minute. Clinical applications such as renography, cardiac output, left ventricular ejection fraction, cerebral blood flow symmetry, regional lung perfusion, etc., can be conveniently accomplished within a few minutes of the patient data acquisition.

RAPID MANIPULATION AND DISPLAY OF COUNTS VS. TIME CURVES
BY HARDWARE AND SOFTWARE

By

Thomas F. Budinger

Donner Laboratory
Lawrence Berkeley Laboratory
University of California

INTRODUCTION

Nuclear medicine quantitation procedures involve three equally important operations: acquisition, data manipulation, and display. Aspects of acquisition and display of two-dimensional images are covered elsewhere in this symposium. This paper covers techniques of rapidly manipulating and displaying curves of counts vs. time. These activity vs. time functions, commonly called uptake curves, area of interest uptake functions (curves), or "rate" of uptake curves, will be referred to here as time functions, TF. Spatial distributions, such as profiles of activity across a crystal or impulse response functions, can be manipulated by the same programs as are used for time functions. Mathematical manipulation of images (frames) is an extension of these techniques.

Some mathematical manipulations of TF's easily implemented on small computer systems are:

1. Addition, subtraction, multiplication and division of TF by scalars.
2. Addition, subtraction, multiplication and division of one TF by another.
3. Differentiation and integration between time or spatial coordinate limits.
4. Modification by logarithmic, exponential, square root and trigonometric operators.
5. Dead-time correction.
6. Convolution of time function with smoothing operator.
7. Fourier or Laplace transform analyses of TF (or spatial functions).
8. Normalization to unit area or volume.

9. Retrospective gating.

The ability to do No. 1 above implies the capability for the other operations; for example, retrospective gating is simply application of differentiation which is based on the ability to subtract one point of a TF from an adjacent datum.

The following operations or studies can be simple procedures of a few minutes for the clinician or researcher who might not have any knowledge of programming:

1. Decay correction (fluorine-18, oxygen-15, nitrogen-13, carbon-11); e.g., quantitation of the turnover of amino acids, carbohydrates and alcohols with camera or scanner.
2. Cross talk correction; e.g., removal of contributed activity from overlapping regions such as pulmonary artery contribution to aortic output curve or Compton addition from dual isotope studies.
3. Left to right cardiac shunt detection; e.g., Jones et al.¹, Alazraki et al.².
4. Renography and computer-assisted blood background subtraction techniques (Britton and Brown³, Secker-Walker et al.⁴).
5. Cardiac compartment turnover transit time studies (Del Guercio⁵, Ishii and MacIntyre⁶, Jones et al.⁷).
6. Left ventricle dV/dT measurements and calculation of ejection fractions (Brooks et al.⁸, Van Dyke et al.⁹).
7. Left ventricular wall movement through retrospective gating.
8. Xenon washout studies.
9. Cerebral blood flow symmetry.
10. Thyroid technetium uptake function.
11. Gall bladder-labelled dye excretion studies.

The procedures listed above and others can be readily effected using small computer systems. The clinician or researcher has the ability to select areas of interest and accumulate time functions such as shown in Fig. 1. The time required for calculating 4 time functions containing 100 to 200 frames should be less than 0.5 min. Rapid and mathematically complex manipulations of time functions are of little use without a convenient means of displaying the results. Thus the initial focus of this paper is on means of displaying scalar functions.

METHODS

Display of time functions and results of arithmetic operations.

Clinically useful and timely presentation of time functions, or the results of operations on time functions such as renograms or dynamic heart and lung studies, is not a trivial problem. In the first place the polaroid images of a digitized function are difficult to read and confusing, because neither overlapping data points describing TF curves or coordinate scale values are

Fig. 1 Renograms generated from Hippuran study using a light pen to designate left and right kidney area of interest for subsequent computer-controlled creation of time functions.

easily identified on the display. Hand labelling is inaccurate and inconvenient. The skilled manipulator can smooth, expand and adjust his display to give a representative picture; e.g., comparison of isotope uptake (Fig. 2). However, these marked polaroid images are hard to catalogue, copy and enter in a patient's chart. Devices such as X-Y plotters, VERSATEC copy system, or the Techtronics hard copy device provide convenient output, but are expensive, require space, and are an added interface to maintain. The simplest and most inexpensive hard copy output device is the teletypewriter printer. One can easily display one to three time functions with selection of scale and time interval. We find the teleprinter paper output curves easy to handle, write on, and color, if desired. A display of two curves of activity recorded at 50 msec time intervals is shown as Fig. 3. The program in BASIC (Hewlett-Packard BASIC) is listed as Appendix I. This program could be written in FOCAL, FORTRAN, NUTRAN, or other languages. It was designed for convenient display of up to 3 TF's for both clinical studies and research. The time functions are read and written by CALL READT (. . . , . . . , . . .) and CALL WRITT (. . . , . . . , . . .) available on the Hewlett-Packard nuclear medicine system. The user can select 1, 2 or 3 time functions to be plotted simultaneously, and has the ability to select portions of the time functions, minimum and maximum values for display, and a list of values. The commands to and queries from the teletype are shown in Appendix II.

Left wall

Right wall

Fig. 2 Comparison of uptake curves in human after 500 μCi Cs-129. Polaroid image is adequate after special labeling and some experience with the display unit.

Fig. 3 Teleprinter output of manipulated time functions. Program is simple and this technique available to any small computer user.

Modes of manipulation.

Three ways whereby time functions can be manipulated are:

- I. Hard-wired program.
- II. Systems program.
- III. Control by software programs in BASIC, FOCAL FORTRAN, etc.

Mode I: The hard-wired system such as might be found in a single or dual parameter analyzer used in spectroscopy and nuclear physics has the capability of TF arithmetic; however, in general it does not have the flexibility of computer-controlled manipulations, and thus without this latter capability Mode I is not acceptable for nuclear medicine applications. The hard-wire approach has speed which becomes important when doing operations such as a Fourier transform. For example, a hard-wired fast Fourier transform system can analyze a data array of 1024 points in 15 milliseconds (Maschke and Cline¹⁰), whereas the computer software controlled approach for only 128 points will take 0.5 min for the fast Fourier transform, and hours for the conventional Fourier transform.

Mode II: The system program mode of operation can be thought of as a computer simulation of hard-wired operations. The commands to the system are via control panel switches, or more conveniently by specified keys of a standard teleprinter or dedicated 3 X 3 keyboard. Thus the keyboard of a teleprinter such as the ASR can serve what appears to be dual functions of usual operation of typed messages to the computer and command code messages to the systems programs. In this mode of control over time functions employed by the Hewlett-Packard scintigraphic system, the ASCII-coded input from the teleprinter is picked up from the I/O bus by the computer executive program (teletype keyboard monitor) and an interpreter transforms the ASCII to an internal system language. Operations such as subtraction of one time function, TF1, from another, TF2, with the result designated another TF, are read as commands in the run buffer. The symbols seen on the teleprinter output are:

TF2 - TF1 > TF3

Subtraction of one renograph from another by this technique is shown in Fig. 1. This mode of TF manipulation is very fast, and is an ideal method of effecting addition, subtraction, multiplication and division of TF's by scalars or by another TF. The system programs under keyboard command can be extended or qualitatively changed without hardware changes. For example, a standard smoothing operation such as the three-point running average is easily implemented as a Mode II procedure. This mode allows the clinician to effect computations on time functions with no need to learn programming skills. The user unfamiliar with a keyboard typewriter can be assisted by coloring the keys in addition to a keyboard label overlay used for marking individual key functions.

Mode III: Control by software programs is the most general mode and allows one to perform all of the possible arithmetic operations so long as sufficient computer memory is available (8K is adequate for operations on 100 to 200 point time functions). Multiple simple operations such as subtraction of time functions after multiplication by scalars can be easily programmed in language appropriate for the system, such as BASIC, FOCAL, NUTRAN, or FORTRAN; however, these simple manipulations can also be performed by programming a sequence of system commands under Mode II operation. Mode III is appropriate for such operations as isotope-decay correction, differentiation, printing a table of ratios, etc. The remainder of this paper is an illustration of some programs appropriate to medical science applications.

Isotope decay correction (Appendix III).

This program calls for the time function to be corrected, the time intervals between points of the time function (between frames), and the half-life of the isotope in minutes. The correction is made by multiplying the time function values at each time by $\exp(t - t_0/T_{1/2})$. The time for execution of this program is of no consequence. The only problem is the logistics of reading in the paper tape, but this can be overcome by having this program assigned to a text buffer and callable from the system's library.

Cross talk correction (Appendix III).

When the right ventricle area overlaps the left ventricle projected area for the anterior view on a cardiac dynamic study, it is desirable to remove from the left ventricle compartment the activity added by the right ventricle. This can be effected by subtracting from the left ventricle time function some percent contribution from the right ventricle. Compton addition from a relatively high energy isotope to the photopeak of a lower energy isotope can be extracted for dual isotope studies. Once again there is no perceptible execution time.

The "honesty" time function.

It is important in evaluating the clinical usefulness of procedures to examine the expected error for a particular study. This is a curve of the expected error in terms of percent. At present the Hewlett-Packard 5407A system has a program for calculating the square root. Rather than write a new Mode III program, the 'honesty' time function can be generated by performing the square root under Mode III, then using Mode II multiply this result by 100, then divide by the original time function. Thus if TF1 is the original data and TF2 the square root, the honesty time function is obtained by--

$$(TF2 * 100)/TF1.$$

Retrospective gating.

It is of some concern that the physiological gates such as from the EKG might not be appropriate gates or signals for selecting desired time intervals when the organ is in a particular functional state. An example is the uncertainty in the time of maximum endiastole. By accumulating all of the data and examining the time functions for maximum counts and minimum counts, the two-dimensional frames that were acquired at those times can be summed and averaged to show the aggregate of endiastole vs. the aggregate exsystole. From such information the aggregate frames can be subtracted, and the ventricle shell representing the ejection volume reconstructed from one view with dangerous assumptions, or two views from a biplanar study. The program to select the frames involves basically normalization followed by differentiation and selection of frames within a window around the zeroes of the derivative function. A program in BASIC performs this operation well, if care is taken to avoid local minima.

Fourier transform.

Signals which are a function of time, and particularly periodic signals such as cardiac dynamic functions, are conveniently analyzed in terms of frequency components. It is relatively easy to implement conventional Fourier and Laplace transform programs on small computers; however, the long time involved in the computation has been restrictive until investigators have applied the fast Fourier transform algorithm (Cochran et al.¹¹, Gentleman and Sande¹²) on both large and small computers (Rothman¹³, Bergland¹⁴, Brigham and Morrow¹⁵, Bice¹⁶, Budinger¹⁷). Once a time function has been transformed to Fourier space, smoothing techniques and Fourier transform analysis methods of biological compartment models are available. Smoothing of time functions by going through Fourier space is not the most practical method in that most smoothing functions can be effected by 3 or 4 point weighted running averages, and convolution of these functions with the time function in real space is fast because the window is small. Programs in BASIC, FOCAL or some other appropriate language are straightforward. Smoothing belongs in the system program, Mode II, operation.

A program for performing the Fourier transform of TF's with 128 or less data points is shown as Appendix IV. This program transforms a 128-point function in 1.2 min. By taking advantage of Friedel's law (neg. frequencies = complex conjugate of pos. frequencies) the computational time is halved. The program was written for research purposes, and is restricted to one dimension. Some clinical applications and interest in two-dimensional image processing by small computers argue for implementation of a hard-wired Fourier analyzer¹⁸ as part of a disc operating system. The feasibility of this approach has been established.

SUMMARY

Of the three basic processes involved in nuclear medicine quantitation (acquisition, manipulation, and display), the manipulation of one-dimensional data arrays representing the counts or activity as a function of time for a selected area of interest is easily handled by small computers. Even Fourier transformation of data arrays of 128 points is feasible. The basic mathematical operations, including exponential and trigonometric functions, can be employed as part of a systems program or user-created software to perform all of the operations that have appeared in the literature or have been suggested for time functions. These operations can be performed by the clinician or researcher who does not have any special skills with computers. Both the mathematical manipulations and display of results can be accomplished with speed and flexibility for such procedures as renography, ejection fraction computation, and even some aspects of noncatenary compartment model calculations. For extension of the one-dimensional time function manipulation procedures to two dimensions, more emphasis is required on hard-wire and assembly language programs so that the time required for operations does not detract from the usefulness and enjoyment of processing nuclear medicine data.

REFERENCES

1. R. H. Jones, P. A. W. Anderson, J. K. Goodrich, and D. C. Sabiston, Jr., "Radionuclide Detection of Left-to-Right Intercardiac Shunts", J. Nucl. Med., 12, 370 (1971).
2. N. P. Alazraki, W. L. Ashburn, A. Hogan, and W. F. Friedman, "Detection of Left-to-Right Cardiac Shunts with the Scintillation Camera Pulmonary Dilution Curve", J. Nucl. Med., 13, 142 (1972).
3. K. E. Britton and N. J. G. Brown, "The Renogram and its Quantization", Brit. J. Urol. Suppl., 41(4), 15 (1969).
4. R. H. Secker-Walker, E. P. Shepherd, and K. J. Cassell, "Clinical Applications of Computer-Assisted Renography", J. Nucl. Med., 13, 235 (1972).
5. L. R. M. Del Guercio, "Contrast Dilution Analysis", Trans.N. Y. Acad. Sci., Series II, 33, 387 (1971).
6. Y. Ishii and W. J. MacIntyre, "Measurement of Heart Chamber Volumes by Analysis of Dilution Curves Simultaneously Recorded by Scintillation Camera", Circulation, 44, 37 (1971).
7. R. H. Jones, R. B. Klaphaak, C. M. Coulam, J. K. Goodrich, and D. C. Sabiston, Jr., "Anatomic Resolution in Dynamic Radionuclide Studies by Computer Identification of Radioactivity Fluctuations with Time", J. Nucl. Med., 12, 371 (1971).
8. R. C. Brooks, V. D. Vandelinde, K. E. Hammermeister, and J. R. Warbasse, "Digital Filtering of Left Ventricular Heart Volume and Calculation of Aortic Valve Blood Flow", Comput. Biomed. Res., 4, 340 (1971).
9. D. C. Van Dyke, H. O. Anger, R. W. Sullivan, Y. Yano, and H. G. Parker, "Cardiac Evaluation from Radioisotope Dynamics", LBL-387 (1971).
10. N. D. Maschke and S. Cline, "New Capabilities in Digital Low Frequency Spectral Analysis", Hewlett-Packard Journal (June 1972)(In press).

11. W. T. Cochran, J. W. Cooley, D. L. Faven, H. D. Helms, R. A. Kaenel, W. W. Lang, G. E. Maling, Jr., D. E. Nelson, C. M. Rader, and P. D. Welch, "What is the Fast Fourier Transform?", IEEE Trans. Audio Electroacoustics, 15, 45 (1967).
12. W. M. Gentleman and G. Sande, in AFIPS Conference Proceedings, Vol. 29, pp. 563-578 (Spartan Books, Washington, 1966).
13. J. Rothman, "The Fast Fourier Transform and its Implementation", Decuscope, 7, 3 (1968).
14. G. D. Bergland, "A Guided Tour of the Fast Fourier Transform", IEEE Spectrum (page 41, July 1969).
15. E. O. Brigham and R. E. Morrow, "The Fast Fourier Transform", IEEE Spectrum, 63 (December 1967).
16. P. K. Bice, "Speed up of the Fast Fourier Transform and Reduce Memory Requirements so that a Minicomputer can do the Whole Job", Electronic Design, 9, 66 (1970).
17. T. F. Budinger, "Transfer Function Theory and Image Evaluation in Biology: Applications in Electron Microscopy and Nuclear Medicine", (Ph.D. Thesis), LBL-565 (1971).

APPENDIX I

PLOT PROGRAM (in BASIC)

```

10  DIM A(160),B(160),C(160)
20  PRINT "*PLOT TIMEFUNCTIONS..."
30  PRINT
40  PRINT "HOW MANY TIMEFUNCTIONS(1<=T<=3)";
50  INPUT T
90  IF T=1 THEN 200
100 IF T=2 THEN 400
110 IF T=3 THEN 600
200  PRINT "WHAT IS THE TIMEFUNCTION";
201  INPUT P
220  PRINT "ENTER NO. POINTS IN TIMEFUNCTION";
225  INPUT Z
230  GOSUB 2000
235  GOSUB 3000
236  PRINT "TABLE OF TF: YES=1 NO=0";
237  INPUT G
238  IF G=0 THEN 240
239  GOSUB 4000
240  GOSUB 2200
241  PRINT
242  PRINT
243  PRINT TAB(31);"TF=";P;"*";
245  GOSUB 2500
250  GOSUB 2600
255  FOR J=M TO N STEP E
260  IF A(J)<7 OR A(J)>71 THEN 280
270  PRINT J;".";TAB(A(J));"*";
275  GOTO 285
280  PRINT J;"."
285  NEXT J
290  PRINT ".....";
295  PRINT ".....";
300  PRINT "DONE"
305  END
400  PRINT "WHAT ARE THE TIMEFUNCTIONS";
401  INPUT P,0
415  PRINT "ENTER NO. POINTS IN TIMEFUNCTIONS";
420  INPUT Z
425  GOSUB 2000
430  GOSUB 3000
435  GOSUB 3100
436  PRINT "TABLE OF TF: YES=1 NO=0";
437  INPUT G
438  IF G=0 THEN 440
439  GOSUB 4000
440  GOSUB 2200
441  PRINT
442  PRINT
443  PRINT TAB(23);"TF=";P;"*";          TF=";Q;"*";
445  GOSUB 2500
450  GOSUB 2600
455  GOSUB 2650
460  FOR J=M TO N STEP E
475  IF (A(J)<7 OR A(J)>71) AND (B(J)<7 OR B(J)>71) THEN 510

```

```

480 IF B[J]<7 OR B[J]>71 THEN 540
485 IF A[J]<7 OR A[J]>71 THEN 550
488 IF A[J]=B[J] THEN 517
490 IF A[J]<B[J] THEN 520
500 IF B[J]<A[J] THEN 530
510 PRINT J; "."
515 GOTO 555
517 PRINT J; ". "; TAB(A[J]); "%"
518 GOTO 555
520 PRINT J; ". "; TAB(A[J]); "% "; TAB(B[J]); "+"
525 GOTO 555
530 PRINT J; ". "; TAB(B[J]); "+ "; TAB(A[J]); "%"
535 GOTO 555
540 PRINT J; ". "; TAB(A[J]); "%"
545 GOTO 555
550 PRINT J; ". "; TAB(B[J]); "+"
555 NEXT J
557 PRINT "          .....";
558 PRINT ".....";
560 PRINT "DONE"
565 END
600 PRINT "WHAT ARE THE TIMEFUNCTIONS";
601 INPUT P,Q,R
615 PRINT "ENTER NO. POINTS IN TIMEFUNCTIONS";
620 INPUT Z
625 GOSUB 2000
630 GOSUB 3000
635 GOSUB 3100
640 GOSUB 3200
641 PRINT "TABLE OF TF; YES=1 NO=0";
642 INPUT G
643 IF G=0 THEN 645
644 GOSUB 4000
645 GOSUB 2200
646 PRINT
647 PRINT
648 PRINT TAB(16); "TF="; P; "%          TF="; Q; "+"          TF="; R; "#"
650 GOSUB 2500
655 GOSUB 2600
660 GOSUB 2650
665 GOSUB 2700
670 FOR J=M TO N STEP E
685 IF (A[J]<7 OR A[J]>71) AND (B[J]<7 OR B[J]>71) THEN 688
687 GOTO 690
688 IF (C[J]<7 OR C[J]>71) THEN 700
690 IF (A[J]<7 OR A[J]>71) AND (B[J]<7 OR B[J]>71) THEN 1080
695 IF (A[J]<7 OR A[J]>71) AND (C[J]<7 OR C[J]>71) THEN 1060
700 IF (B[J]<7 OR B[J]>71) AND (C[J]<7 OR C[J]>71) THEN 1040
702 IF (C[J]<7 OR C[J]>71) AND A[J]=B[J] THEN 1055
705 IF (C[J]<7 OR C[J]>71) AND A[J]<B[J] THEN 920
710 IF (C[J]<7 OR C[J]>71) THEN 940
712 IF (B[J]<7 OR B[J]>71) AND A[J]=C[J] THEN 1055
715 IF (B[J]<7 OR B[J]>71) AND A[J]<C[J] THEN 960
720 IF (B[J]<7 OR B[J]>71) THEN 980
722 IF (A[J]<7 OR A[J]>71) AND B[J]=C[J] THEN 1075
725 IF (A[J]<7 OR A[J]>71) AND B[J]<C[J] THEN 1000
730 IF (A[J]<7 OR A[J]>71) THEN 1020
735 IF A[J]<B[J] AND B[J]<C[J] THEN 800
737 IF A[J]<B[J] AND B[J]=C[J] THEN 935
739 IF A[J]=C[J] AND C[J]<B[J] THEN 1035

```

```

740 IF A[J]<C[J] AND C[J]<E[J] THEN 820
742 IF B[J]<A[J] AND A[J]<C[J] THEN 840
745 IF B[J]<A[J] AND A[J]=C[J] THEN 955
747 IF B[J]<C[J] AND C[J]<A[J] THEN 860
748 IF E[J]=C[J] AND C[J]<A[J] THEN 975
750 IF C[J]<A[J] AND A[J]<B[J] THEN 880
752 IF C[J]<A[J] AND A[J]=B[J] THEN 995
755 IF C[J]<B[J] AND B[J]<A[J] THEN 960
757 IF A[J]=B[J] AND B[J]<C[J] THEN 1015
760 IF A[J]=B[J] AND B[J]=C[J] THEN 1050
760 PRINT J;". "
790 GOTO 1090
800 PRINT J;". ";TAB(A[J]);"*";TAB(B[J]);"+";TAB(C[J]);"#"
810 GOTO 1090
820 PRINT J;". ";TAB(A[J]);"*";TAB(C[J]);"#" ;TAB(E[J]);"+ "
830 GOTO 1090
840 PRINT J;". ";TAB(B[J]);"+";TAB(A[J]);"*";TAB(C[J]);"#"
850 GOTO 1090
860 PRINT J;". ";TAB(B[J]);"+";TAB(C[J]);"#" ;TAB(A[J]);"*"
870 GOTO 1090
880 PRINT J;". ";TAB(C[J]);"#" ;TAB(A[J]);"*";TAB(B[J]);"+ "
890 GOTO 1090
900 PRINT J;". ";TAB(C[J]);"#" ;TAB(E[J]);"+";TAB(A[J]);"*"
910 GOTO 1090
920 PRINT J;". ";TAB(A[J]);"*";TAB(B[J]);"+ "
930 GOTO 1090
935 PRINT J;". ";TAB(A[J]);"*";TAB(B[J]);"X"
936 GOTO 1090
940 PRINT J;". ";TAB(B[J]);"+";TAB(A[J]);"*"
950 GOTO 1090
955 PRINT J;". ";TAB(B[J]);"+";TAB(A[J]);"X"
956 GOTO 1090
960 PRINT J;". ";TAB(A[J]);"*";TAB(C[J]);"#"
970 GOTO 1090
975 PRINT J;". ";TAB(B[J]);"X";TAB(A[J]);"*"
976 GOTO 1090
980 PRINT J;". ";TAB(C[J]);"#" ;TAB(A[J]);"*"
990 GOTO 1090
995 PRINT J;". ";TAB(C[J]);"#" ;TAB(A[J]);"X"
996 GOTO 1090
1000 PRINT J;". ";TAB(B[J]);"+";TAB(C[J]);"#"
1010 GOTO 1090
1015 PRINT J;". ";TAB(A[J]);"X";TAB(C[J]);"#"
1016 GOTO 1090
1020 PRINT J;". ";TAB(C[J]);"#" ;TAB(B[J]);"+ "
1030 GOTO 1090
1035 PRINT J;". ";TAB(A[J]);"X";TAB(B[J]);"+ "
1036 GOTO 1090
1040 PRINT J;". ";TAB(A[J]);"*"
1050 GOTO 1090
1055 PRINT J;". ";TAB(A[J]);"X"
1056 GOTO 1090
1060 PRINT J;". ";TAB(B[J]);"+ "
1070 GOTO 1090
1075 PRINT J;". ";TAB(B[J]);"X"
1076 GOTO 1090
1080 PRINT J;". ";TAB(C[J]);"#"
1090 NEXT J
1092 PRINT "
1093 PRINT "....."

```

```

1095 PRINT "DONE"
1098 END
2000 LET I=1
2010 CALL READT(P,Z,A(I))
2020 IF T-I=0 THEN 2110
2030 LET I=I+1
2040 CALL READT(G,Z,B(I))
2050 IF T-I=0 THEN 2110
2060 LET I=I+1
2070 CALL READT(R,Z,C(I))
2110 RETURN
2200 PRINT "MIN, MAX ORDINATE FOR PLOT:";
2205 INPUT K,L
2210 IF K<0 THEN 2200
2215 IF K>L THEN 2200
2220 PRINT "MIN, MAX FRAME:";
2225 INPUT M,N
2230 IF M<1 THEN 2220
2235 IF M>Z THEN 2220
2240 IF N<M THEN 2220
2245 IF N>Z THEN 2220
2250 PRINT "INCHMENT:";
2255 INPUT E
2260 IF E<1 THEN 2255
2265 IF E>Z THEN 2255
2270 RETURN
2500 PRINT
2510 PRINT
2520 PRINT TAB(6);K;TAB(35);"COUNTS";TAB(65);L
2530 PRINT "FRAME .....";
2540 PRINT ".....";
2550 RETURN
2600 FOR J=M TO N STEP E
2610 LET A(J)=(A(J)-K)/(L-K)
2620 LET A(J)=INT(A(J)*64+7)
2630 NEXT J
2640 RETURN
2650 FOR J=M TO N STEP E
2660 LET B(J)=(B(J)-K)/(L-K)
2670 LET B(J)=INT(B(J)*64+7)
2680 NEXT J
2690 RETURN
2700 FOR J=M TO N STEP E
2710 LET C(J)=(C(J)-K)/(L-K)
2720 LET C(J)=INT(C(J)*64+7)
2730 NEXT J
2740 RETURN
3000 LET V=0
3010 FOR J=1 TO Z
3020 IF A(J)>V THEN 3040
3030 GOTO 3050
3040 LET V=A(J)
3050 NEXT J
3060 PRINT "MAXIMUM FOR TF=";P;"IS",V
3070 RETURN
3100 LET V=0
3110 FOR J=1 TO Z
3120 IF B(J)>V THEN 3140
3130 GOTO 3150
3140 LET V=B(J)

```

```

3150 NEXT J
3160 PRINT "MAXIMUM FOR TF="; Q; "I S", V
3170 RETURN
3200 LET V=0
3210 FOR J=1 TO Z
3220 IF C[J]>V THEN 3240
3230 GOTO 3250
3240 LET V=C[J]
3250 NEXT J
3260 PRINT "MAXIMUM FOR TF="; R; "I S", V
3270 RETURN
4000 PRINT
4005 PRINT "TABLE FOR TIMEFUNCTIONS"
4010 PRINT "MIN, MAX FRAME:";
4015 INPUT M,N
4020 PRINT "INCREMENT:";
4025 INPUT E
4030 IF T=2 THEN 4050
4035 IF T=3 THEN 4070
4040 PRINT "-----"
4041 PRINT TAB(15); "FRAME"; TAB(30); "TF="; P
4042 FOR I=M TO N STEP E
4044 PRINT TAB(15); I; TAB(30); A[I]
4046 NEXT I
4047 PRINT "-----"
4048 GOTO 4090
4050 PRINT "-----"
4051 PRINT TAB(10); "FRAME"; TAB(25); "TF="; P; TAB(35); "TF="; Q
4052 FOR I=M TO N STEP E
4054 PRINT TAB(10); I; TAB(25); A[I]; TAB(35); B[I]
4056 NEXT I
4057 PRINT "-----"
4058 GOTO 4090
4070 PRINT "-----"
4071 PRINT TAB(5); "FRAME"; TAB(20); "TF="; P; TAB(30); "TF="; Q;
4072 PRINT TAB(40); "TF="; R
4073 FOR I=M TO N STEP E
4074 PRINT TAB(5); I; TAB(20); A[I]; TAB(30); B[I]; TAB(40); C[I]
4076 NEXT I
4077 PRINT "-----"
4078 GOTO 4090
4090 RETURN
5000 END

```

APPENDIX II

Queries and Commands for PLOT & FOURIER TRANSFORM Programs.

RUN

*PLOT TIMEFUNCTIONS...

HOW MANY TIMEFUNCTIONS(1<=T<=3)?3

WHAT ARE THE TIMEFUNCTIONS?1,2,3

ENTER NO. POINTS IN TIMEFUNCTIONS?160

MAXIMUM FOR TF= 1   IS   2943

MAXIMUM FOR TF= 2   IS   1471

MAXIMUM FOR TF= 3   IS   1476

TABLE OF TF; YES=1 NO=0?1

TABLE FOR TIMEFUNCTIONS

MIN, MAX FRAME: ?1,30

INCREMENT: ?2

FRAME	TF= 1	TF= 2	TF= 3
1	332	441	446
3	2461	1030	1035
5	2676	1338	1343
7	2913	1459	1464
9	2921	1460	1465
11	2731	1390	1395
13	2559	1279	1284
15	2300	1150	1155
17	2030	1015	1020
19	1767	883	888
21	1521	760	765
23	1297	648	653
25	1098	549	554
27	923	461	466
29	772	386	391

MIN, MAX ORDINATE FOR PLOT: ?100,2000

MIN, MAX FRAME: ?1,60

INCREMENT: ?2

FIN

***** FOURIER TRANSFORM *****

FOURIER TRANSFORM ENTER 1; INVERSE TRANSFORM ENTER 0?1
DATA ARRAY IS WHAT POWER OF 2??

FUNCTION TO BE TRANSFORMED, TF=?1

WAIT

DONE

WANT A LISTING?(YES=1;NO=0)?0
TO TRANSFER MODULUS TO TF ENTER 1
TO CHANGE VALUES OF N AND S ARRAYS ENTER 2
TO EXECUTE INVERSE TRANSFORM ENTER 3
TO EXECUTE FORWARD TRANSFORM ENTER 5
TO EXIT PROGRAM ENTER 6
?6

READY

PRINT

APPENDIX III ISOTOPE DECAY AND CROSS TALK CORRECTIONS

```

15 DIM A(160),B(160),C(160),D(160)
20 PRINT "ISOTOPE PHYSICAL DECAY CORRECTION, ENTER 1";
25 PRINT ""
30 PRINT "TIME FUNCTION CORRECTION FOR CROSS TALK, ENTER 2"
40 PRINT "YOUR DECISION";
60 INPUT T
70 IF T=1 THEN 100
80 IF T=2 THEN 500
100 PRINT "TIME FUNCTION TO BE CORRECTED; TF=";
102 INPUT TF
103 PRINT "NO. POINTS IN TIME FUNCTION=";
105 INPUT Z
106 PRINT "TIME PER FRAME IN MINUTES";
107 INPUT C
108 PRINT "HALF-LIFE IN MINUTES";
109 INPUT V
110 CALL READT(T,Z,A())
120 LET D=0
125 FOR I=1 TO Z
127 LET F=EXP(D/V)
130 LET B(I)=A(I)*F
135 LET D=D+0
139 NEXT I
140 PRINT "DONE"
150 PRINT "TRANSFER TO TF=";
155 INPUT F
160 CALL WPITT(F,Z,B())
170 PRINT "DONE"
180 PRINT
190 PRINT "TO DISPLAY THE DIFFERENCE, ENTER 1";
215 INPUT E
217 IF E=1 THEN 220
218 IF E=0 THEN 330
220 FOR I=1 TO Z
230 LET C(I)=B(I)-A(I)
240 NEXT I
250 PRINT "TRANSFER DIFFERENCE TO TF=";
255 INPUT F
260 CALL WRITT(F,Z,C())
330 PRINT "TO EXIT BASIC, ENTER 1";
331 INPUT F
332 IF F=1 THEN 340
333 GOTO 341
340 CALL DONE
341 END
500 PRINT "TIME FUNCTION TO BE CORRECTED IS";
510 INPUT D
520 PRINT "TIME FUNCTION CONTRIBUTING CROSS TALK IS";
530 INPUT R
540 PRINT "PERCENT OF THAT FUNCTION WHICH IS CROSS TALK IS";
550 INPUT S
551 LET S=S/100
560 PRINT "NUMBER OF DATA POINTS IS";
570 INPUT Z
580 CALL READT(D,Z,A())
590 CALL READT(R,Z,B())
600 FOR I=1 TO Z
610 LET B(I)=B(I)*S
620 NEXT I
630 FOR I=1 TO Z
640 LET C(I)=A(I)-B(I)
650 NEXT I
670 PRINT "DONE"
680 PRINT "PUT RESULT IN TF=";
690 INPUT F
700 CALL WPITT(F,Z,C())
900 END

```

APPENDIX IV

```

1 PRINT "          ***** FOURIER TRANSFORM *****"
2 PRINT
3 PRINT "FOURIER TRANSFORM ENTER 1; INVERSE TRANSFORM ENTER 0";
5 INPUT E
10 DIM R(128),S(128),A(128)
20 PRINT "DATA ARRAY IS WHAT POWER OF 2";
21 INPUT P
23 LET N=2↑P
24 LET M=N
25 LET N=2↑P
27 LET C=8*ATN(1)/N
28 IF E=0 THEN 800
29 PRINT
30 PRINT " FUNCTION TO BE TRANSFORMED, TF=";
31 INPUT T
32 CALL READT(T,M,A(1))
50 FOR I=1 TO N
51 LET R(I)=A(I)
52 LET S(I)=0
54 NEXT I
70 IF E=0 THEN 100
75 LET E=1
80 FOR I=1 TO N
82 LET R(I)=R(I)/N
86 NEXT I
100 FOR L=0 TO P-1
110 LET P1=2↑(P-L-1)
120 LET M=0
130 FOR I=1 TO 2↑L
140 LET K1=INT(M/P1)
150 GOSUB 780
160 LET Y1=COS(C*K2)
170 LET Y2=SIN(C*K2)
171 IF E=0 THEN 180
172 LET Y2=-Y2
180 FOR J=1 TO P1
190 LET Y3=R(M+P1+1)*Y1-S(M+P1+1)*Y2
200 LET Y4=R(M+P1+1)*Y2+S(M+P1+1)*Y1
210 LET R(M+P1+1)=R(M+1)-Y3
220 LET S(M+P1+1)=S(M+1)-Y4
230 LET R(M+1)=R(M+1)+Y3
240 LET S(M+1)=S(M+1)+Y4
250 LET M=M+1
260 NEXT J
270 LET M=M+P1
280 NEXT I
290 NEXT L
291 PRINT "WAIT"
300 FOR I=0 TO N-1
310 LET K1=I
320 GOSUB 780
330 IF K2>1 THEN 390
340 LET K3=R(I+1)
350 LET R(I+1)=R(K2+1)
360 LET R(K2+1)=K3
365 LET K3=S(I+1)
370 LET S(I+1)=S(K2+1)
380 LET S(K2+1)=K3
390 NEXT I
395 GOSUB 850

```



```

399 PRINT "DONE"
400 PRINT
401 PRINT "    WANT A LISTING?(YES=1;NO=0)";
410 INPUT D
411 IF D=1 THEN 600
415 GOTO 650
600 PRINT
601 PRINT
602 PRINT "                REAL PART    IMAGINARY PT.    MODULUS"
610 PRINT
620 FOR I=0 TO N-1
630 PRINT I,R[I+1],S[I+1],A[I+1]
640 NEXT I
650 PRINT "TO TRANSFER MODULUS TO TF ENTER 1"
651 PRINT "TO CHANGE VALUES OF R AND S ARRAYS ENTER 2"
652 PRINT "TO EXECUTE INVERSE TRANSFORM ENTER 3"
653 PRINT "TO EXECUTE FORWARD TRANSFORM ENTER 5"
654 PRINT "TO EXIT PROGRAM ENTER 6"
655 INPUT V
660 IF V=1 THEN 700
661 IF V=2 THEN 1000
662 IF V=3 THEN 100
663 IF V=4 THEN 100
664 IF V=5 THEN 75
665 IF V=6 THEN 2000
700 PRINT "TRANSFER MODULUS TO TF=";
705 INPUT T
706 CALL WRITT(T,N,R[1])
770 GOTO 651
780 LET K2=0
790 FOR K=1 TO P
792 LET K3=K1-2*INT(K1/2)
794 LET K1=INT(K1/2)
796 IF K3=0 THEN 798
797 LET K2=K2+2*(P-K)
798 NEXT K
799 RETURN
800 PRINT
801 PRINT "%INPUT REAL AND IMAGINARY VALUES FOR INVERSE TRANSFORM"
802 FOR I=1 TO N
803 PRINT "R,S";
804 INPUT R[I],S[I]
805 NEXT I
806 GOTO 100
850 FOR I=1 TO N
860 LET A[I]=(S[I]*2+R[I]*2)
870 LET A[I]=A[I]*.5
880 NEXT I
885 RETURN
1000 PRINT
1010 PRINT " FOR LIST OF MODULUS ARRAY INPUT 1";
1011 INPUT U
1020 IF U=1 THEN 1100
1021 PRINT "WANT TO ZERO COMPONENTS(YES=1;NO=0)";
1022 INPUT X
1023 IF X=1 THEN 1200
1024 GOTO 650
1100 PRINT "    ***** A  ARRAY*****"
1101 PRINT

```

```

1109  FOR I=1 TO N
1110  PRINT I,A[I]
1115  NEXT I
1120  GOTO 1021
1200  PRINT
1201  PRINT 'MIN,MAX PTS. TO ZERO';
1202  INPUT M,N
1210  FOR I=M TO N
1220  LET R[I]=0
1230  LET S[I]=0
1240  NEXT I
1250  PRINT " INVERSE TRANSFORM INPUT 1";
1251  INPUT B
1252  IF B=1 THEN 100
1260  GOTO 650
1300  PRINT " LIST OF MODIFIED DATA INPUT 1";
1301  INPUT C
1302  IF C=1 THEN 1400
1303  GOTO 650
1400  PRINT "          ***** LIST OF MODIFIED DATA *****"
1401  PRINT
1410  FOR I=1 TO N
1420  PRINT R[I],S[I]
1430  NEXT I
1440  GOTO 650
2000  END

```

SOME GUIDELINES FOR DOCUMENTATION PRACTICES
AS AN AID IN THE SHARING OF
COMPUTER PROGRAMS AND TECHNOLOGY

Betty F. Maskewitz (Radiation Shielding Information Center)
W. J. McClain (Computing Technology Center)

Oak Ridge National Laboratory
Oak Ridge, Tennessee 37830

April 1972

SOME GUIDELINES FOR DOCUMENTATION PRACTICES
AS AN AID IN THE SHARING OF
COMPUTER PROGRAMS AND TECHNOLOGY

Betty F. Maskewitz (Radiation Shielding Information Center)
W. J. McClain (Computing Technology Center)
Oak Ridge National Laboratory

ABSTRACT

Documentation is an integral part of computer code development. Effective utilization of a program depends on the availability of a written description which includes a user's manual. Guidelines are proposed as an aid in such documentation. The computer code abstract, application information, the problem or function definition, programming information, and the transmittal package are each covered in the proposed guidelines.

Introduction

Documentation is an integral part of computer code development, a normal part of its cost. Effective utilization of a computer code hinges on documentation which describes it and presents instructions on how to use it. While unnecessary and unreasonable burden to the originator must be avoided, his task is incomplete until the program has been properly documented.

Computer codes should be developed in anticipation of wide use, and obvious obstacles to transfer onto other computer systems should be avoided. However, the originator should not be burdened with documenting requirements for implementing a program on distant facilities with which he is not familiar. Documentation must be relevant to a particular version of a program. If there is active development, the documentation should be upgraded as necessary - looseleaf, for easy revision, is a practical way to do it. Each individual documenting report should stand alone and should include a concise abstract of its contents. Use of published material and ideas originated by others should be acknowledged. Only readily available publications are suitable as references.

Technological advances are necessary in the field of computation; therefore, nothing in these guidelines is intended to inhibit development and experimentation.

The scope of the guidelines covers the documentation of digital computer codes designed for applications in nuclear medicine. The objective of the guidelines is to facilitate effective usage, interchange, modification, and extension of such programs.

The guidelines cover five categories, each an essential part of the documentation. These are:

- A. Computer Code Abstract - a brief description of the program;
- B. Application Information - documentation directed to the program use (i.e., a user's manual);
- C. Problem or Function Definition - a formalization of the problem, the theoretical foundation, and solution algorithms;
- D. Programming Information - detailed information regarding program implementation;
- E. Transmittal Package - the material necessary for transfer of the computer program.

A. Computer Code Abstract

The Computer Code Abstract provides a summary of the capabilities of the program and the requirements for its implementation. The abstract should convey sufficient information to permit the reader to assess the applicability of the program to his needs and to estimate the effort required to make it operational. The format given here should be followed to reap the benefits of a single form.

- 1. Name and Descriptive Title: Identify the program and provide any pertinent discussion and acknowledgements.
- 2. Computer Programming Language: Identify the hardware on which the program has been run and give the percentages of each different programming language used.
- 3. Function: Describe briefly the problems solved and/or functions of the program.
- 4. Method of Solution: Describe the primary techniques and major algorithms employed for solution and indicate capabilities and unusual features.
- 5. Restrictions: Discuss limitations associated with usual program application, as dependence of capability on available computer facilities.
- 6. Running Time: Provide information for a user to estimate computer time and cost in usual application.
- 7. Related Programs: List any programs used with or auxiliary to this program, especially those directly coupled through interface data files.
- 8. Hardware Requirements: List the computer hardware required for full utilization of the program, e.g., the amount of memory, disk and drum storage, number of tapes, channel configuration, and input/output equipment.
- 9. System: Identify the monitor and/or the operating system used and the assembler/compiler version.
- 10. Authors: Give the names and addresses of the authors and designate someone as currently carrying primary responsibility for the program.

11. Material Available: Describe the contents of the transmittal package and tell how to obtain it and from where, and note any restrictions or changes.
12. Category and Keywords: List category and keywords to be used for classification for possible information retrieval or cataloging.
13. References.

B. Application Information

This information is directed to the needs of the program user. It should be sufficiently detailed to permit effective use of the program to the extent of serving as a referral document for preparation of input data and interpretation of problem results.

1. Program Description. This narrative should convey the nature of the problem solved or should define the processing tasks performed, explaining the methods and procedures employed. A schematic or logical chart may be necessary to display the flow of the program.
2. Program Considerations. The following important factors should be discussed to permit a program to be usefully applied:
  - a. The major program options that are available to the user.
  - b. Alternate paths which may be selected by the program. For example, an alternate path may be selected if an iterative process fails to satisfy specified convergence criteria.
  - c. The dimensional limitations for major arrays.
  - d. The dependence of computer execution time on the problem dimensions, the paths of computation, and the specified input variables.
  - e. The range limitations for major variables.
  - f. The units of measurement required to produce meaningful solutions from the implemented equations. Note the associated input data requirements and interpretation of results.
  - g. The values assigned to constants built into the program; for example, a table entry might be:

Avogadro's number 0.60225×10^{24} particles/gram-mole
  - h. Restart and recovery procedures from the user's viewpoint.
  - i. Any man-machine interaction considerations.
  - j. Special forms of output available (e.g., microfilm, microfiche, cathode ray tube display, plotters).
  - k. The use and maintenance of data libraries.

3. External Data Files and Related Programs. External data files are all those files which exist prior to execution of and after completion of the computer run. Items that should be included are:
  - a. The use of input data files and the generation of output data files.
  - b. The structure of each external data file.
  - c. The relationship of each external data file to the execution of the program.
  - d. Related programs which generate or process external data files if pertinent to the application of this program.
4. Input Data Requirements. This description should provide the detailed input for execution of the program.
  - a. General information.
    - (1) Describe type and format specifications.
    - (2) Describe special input techniques used for processing data arrays.
    - (3) Identify separation of individual problems in a single run and other blocking of input data.
    - (4) Discuss pitfalls associated with input data as found in typical use.
  - b. Additional special information required.
    - (1) The function of each major control parameter and the specific options available.
    - (2) The related or combined effect of the major control options which are not independent, especially any combinations which over-determine the problem.
    - (3) Error stop information given in edits and specific stops related to causes and to locations in the subprograms.
5. System Control Requirements. Describe any operating system control requirements, indicating interdependence with input options and data files.
6. Output Information. Describe the form and complete contents of the program output. If the output is a picture displayed on a scope or a listed map of activity, give an example of the results.
7. Sample Problem. Provide a representative sample problem(s) when feasible. Describe the physical problem and auxiliary data files. Present the associated input data and discuss these and the results.
  - a. To best serve its intended function, a sample problem and its execution should meet the following requirements:

- (1) It should be drawn from a reference or benchmark situation well known in the field of effort when possible, even though elaboration may be necessary for it to serve adequately.
  - (2) It should exercise a large fraction of the available programmed options, yet take only a reasonable amount of computer time.
  - (3) All of the hardware normally used by the program should be used.
- b. Discuss any special aspects of the execution run.
  - c. Report the expected computer execution time for the sample problem(s).

C. Problem or Function Definition

Background is provided by defining the problems solved, describing the physical, mathematical, or other model employed, and documenting the computational algorithms and numerical techniques implemented in the program. The intent must be to convey a thorough understanding of the foundation of the program. The literature may be referenced, if necessary, to satisfy this requirement.

1. Description. A comprehensive description of the problem solved or of the data processing functions performed is required. The description of the physical theory in terms of a model needs to be reasonably self-contained, referencing sources of the model and of the numerical techniques involved. Sufficient detail is needed to permit a user to judge the suitability of the model for application to a particular situation. Assumptions should be noted and information given about limitations.
2. Data Libraries. Background information on the contents and the use of these data should be provided.
3. Procedures which may be followed to establish the reliability of results obtained in a specific application should be described.

D. Programming Information

This information is directed to the individual responsible for implementing the program on his computer, modifying or extending it to meet local needs, or converting it to the local environment and mode of operation.

1. The Source Program. A properly commented source program may be listed as computer-produced documentation. In addition, supplementary information about the program structure and the individual subprogram is needed. It is helpful if the author should pinpoint known areas of dependency upon the computer system or the local installation support facilities.
  - a. Program structure and subprogram details.
    - (1) Identify the source language(s) in which the program is written.
    - (2) Define the role and function of the main program and each subprogram, its formal parameters and its use, what subprograms call this subprogram or are called by it, which formal parameters are passed and in which direction, and

where they are modified.

- (3) Include a schematic flowchart showing the overall program structure and logic, and also detailed flowcharts where appropriate. The subprogram names should be referenced.
- (4) Relate the problem variables and constants to the programmed symbols. Define the physical units in which they are specified, so that the relationship between these quantities and the problem solved (or the function performed) is apparent.
- (5) Describe shared storage assignments when used.
- (6) Describe in detail the functions performed by machine-dependent subprograms; if possible, provide a high level language equivalent of such routines, preferably FORTRAN.

b. Troublesome areas.

- (1) Identify the use of features unique to a particular language implementation or system; for example, does performance depend on any initial state of the memory?
- (2) Discuss known areas of conversion difficulty and give locations in the subprograms where there may be trouble. Include discussion of such items as data-type equivalences, bit and character manipulation, numerical and character packing and unpacking, and word-length dependencies of constants or variables.
- (3) Enumerate known difficulties or diagnostics associated with assembly or compilation of the source language. Note when special action had to be taken to bypass errors.
- (4) Report the extent to which the programming has been acceptance tested or checked out.

2. Program Implementation Material. This is a description of the environment in which the program is executed. It also provides a useful background for program transfer from the originating installation to another.

a. Hardware requirements.

- (1) List the machine configuration(s) on which the program has been tested successfully.
- (2) Enumerate the main memory storage requirements, the amount and type of auxiliary storage (such as drum, disk, data cell, and tapes) used, and the peripheral equipment (i.e., punch, printer, plotter) needed.
- (3) Describe any special hardware utilized (clock, on-line communication channel, etc.); any hardware interface (a schematic, if possible).

b. Software requirements.

- (1) Identify the operating system and associated subroutine libraries invoked by the program. If these are generally provided by the computer manufacturer or associated software agency, merely cite the appropriate versions and releases and the reference literature.
- (2) Describe all deviations from the manufacturer's supported software. Provide references to generally available documentation of any mathematical or utility library routines supplied locally, installation-dependent features of the operating system, etc.

c. Program considerations.

- (1) Explain the system control commands (cards or statements) required to execute the program.
- (2) Explain procedures used to modify the program at run time.
- (3) Document the overlay or segmentation scheme, if one is used.
- (4) Describe the storage allocation and data management procedures. Indicate the problem-dependent nature of the memory requirements. Discuss program alternatives which affect data storage and use of data buffering.
- (5) Enumerate the logical devices employed, and describe the use of each unit and any data blocking schemes associated with this use. Give rationale for these choices. Define the contents and format of the information resident on each device, its logical designation and its use, where it is altered and accessed and under what circumstances. Discuss related physical device use and requirements.
- (6) Outline the restart, recovery, and successive case capability. Identify data file retention and allocation requirements, next case versus initial case input data specifications and associated limitations and default conventions.
- (7) Discuss program procedures related to the use and maintenance of associated data libraries, files, or collections.

3. External Considerations. A program is often developed under constraints invoked by its destined use as one of a set of programs or as a module in a larger system. These constraints must be understood by a person implementing it at another installation.

- a. Supply a description of the constraints imposed by membership in the larger system. Describe the function of this program in the system, as well as the functions of related member programs, or modules, which normally interact with this one.
- (1) Describe calculational or other paths which incorporate this program.
  - (2) Define the interfaces of this program with other members of the system.

- b. Show, by example, the role this program might be expected to play in an integrated collection of programs indicating interfaces and the communication paths required to effect this integration.

E. Transmittal Package

A transmittal package for a program is the aggregate of card decks, tapes, and documents necessary for its use at another installation.

A short program under 500 cards may be transmitted in card form or as a computer listing of the source program.

Usual transmission of a code is on punched-paper or magnetic tape. A label attached to the tape reel should show how the tape was written (number of tracks, density, character mode, parity, blocking factor and any record or file or end-of-tape sentinels, or whatever applies).

The items in a transmittal package should be listed in a cover letter. This letter should describe the contents of all tapes in the order in which the files are written and repeat the information given on the tape reels about how they were written. Typical contents of a transmittal package follow:

1. Program documentation corresponding to the version of the program included.
2. Complete source program.
3. Library data files required as input to the program.
4. Input data for the sample problem(s).
5. System control cards for the execution of the program on the originator's computer, including allocation of main core, logical devices, and overlay structure, when appropriate.

CODE CONVERSION BY COMPUTER

A. E. McDow, Jr.

April 22, 1972

OAK RIDGE ASSOCIATED UNIVERSITIES

MEDICAL DIVISION

CODE CONVERSION BY COMPUTER

ABSTRACT

Sharing computer programs among participants in the nuclear medicine field is a desirable interaction. However, due to non-standard hardware and software conversions, the sharing process is hampered. An effort to reduce some of the problems involved in sharing programs will be explored with reference to a computer program which does code conversion.

It is possible to write a computer conversion program which can assist a programmer in the conversion process; however, a study of the problems involved indicates that better program documentation is the best solution.

INTRODUCTION

Before we can take a computer program written for a particular machine type and configuration and make it operational on a different machine, we must consider the compatability of the operating systems, machine designs, and machine configurations as well as the operating efficiency of the final converted program.^{1,2,3} In an effort to explore these areas of interest, three computers (PDP-8, PDP-11, and IBM 1800) were chosen to test the feasibility of a computer code conversion program. For the sake of convenience the IBM 1800 was chosen as the final object machine.

DISCUSSION

Perhaps the best avenue of attack for the conversion process by computer is to take a simple problem statement coded for one machine and attempt a conversion to other machines and then explore the difficulties encountered. The statement of the problem¹ (Fig 1) can be expanded into a flow chart (Fig 2) to assist us in understanding the logical steps in the problem solution. Let us arbitrarily assume that the original program is coded for the PDP-8 and is to be converted to operate on the PDP-11 and IBM 1800.

If we assume that the resulting assembler code for each machine can be properly loaded (either as an absolute or as a relocatable module), we can reduce the influence of the operating system on the conversion process. Since there is adequate core space and since no input/output or interrupt generation is involved, the problem is reduced to a matter of code conversion⁴ by a macro expansion scheme for all three basic machines.

GIVEN 100 POSITIVE NUMBERS LESS THAN 20 STORED AT LOCATIONS
DATA +0, +1, +2, ----, +99.

ADD THE NUMBERS TOGETHER AND STORE IN LOCATION SUM.

FIGURE 1. SAMPLE PROBLEM DEFINITION.¹

FIGURE 2. FLOW CHART OF THE SAMPLE PROBLEM.¹

Examination of the PDP-11 addressing scheme reveals a serious inconsistency between the core-to-core reference and the core-to-accumulator type reference used in the PDP-8 and IBM 1800. For example, if a program attempts to use direct address calculation, the PDP-11 code generated would produce errors (Fig 3). The INC ADD command instead of incrementing the address of DATA changes the ADD instruction address-field. When this occurs further conversion of PDP-8 to PDP-11 code is discontinued. With the exception of the word lengths (11 bits on the PDP-8 and 16 bits on the IBM 1800), the PDP-8 and IBM 1800 are the most compatible machines for code conversion. Having made some limiting assumptions and ruling out the worst case conversion (PDP-11), we are left with the best environment and conditions to test our conversion program. If the efficiency of the converted program is acceptable, one could conclude that for at least two machines computer-controlled code conversion would be feasible.

The converted program for the IBM 1800 (Fig 3) will operate satisfactorily but will require additional core locations and execution time. Even the best machine conversion (PDP-8 to IBM 1800) poses some problems that require the attention of a programmer to insure that the intended operations of certain types of instructions were properly converted. For example the word-size difference doesn't present a problem for shift operation in arithmetical operations but byte shifting can create errors in logic.

If one reconsiders the problem definition and flow chart, and codes the problem directly into IBM 1800 Assembler, a much more efficient program is obtained (Fig 4). This procedure allows the programmer to take advantage of a particular machine's features. The directly generated program runs in less time and requires less core than the original PDP-8 program. This example should not be considered an endorsement of any machine but instead a demonstration of the variability of programmers and the complex problems involved in a simple conversion attempt.

CONCLUSION

To remove the limiting assumptions of the conversion of computer programs by a computer program is to open Pandora's Box. The simple operating system now becomes a hostile environment for the converted program. Machine configurations (core size and input/output equipment) require modification of the converted program. Some machines because of their architecture make efficient direct translation impractical.

As this simple example shows, a good definition of the problem and its solution can provide a sound means of sharing programs between users with different computer configurations. While complete documentation and flow charting seems to be an undesirable procedure for many programmers, those programs that need to be shared require good documentation.

PDP 8				IBM 1800		COMMENTS	PDP-11	
CLA	CLL			SLA	17	CLEAR AC AND OVERFLOW	CLR	AC
TAD	K100			A	K100	LOAD 100	CLC	
CIA				EOR	BFFFF	NEGATE COUNT	CLV	
				A	ONE		ADD	K100,AC
DCA	TALLY			STO	TALLY	STORE IN TALLY	NEG	AC
				SLA	16	ZERO AC	MOV	AC,TALLY
ADD,	TAD	DATA	ADD	A	DATA	SUM DATA POINTS	ADD1;	ADD DATA,AC
ISZ	ADD			MDX	L ADD,1	INCREMENT ADDRESS	INC	ADD1
ISZ	TALLY			MDX	L TALLY,1		***ASSEMBLY DISCONTINUED	
JMP	ADD			MDX	ADD	NOT DONE LOOP		
DCA	SUM			STO	SUM	FINIS STORE SUM		
				SLA	16	ZERO AC		
HLT				WAIT		REST AWHILE		
K100,	0100			K100	DC	100	DATA AREA	
TALLY,	0000			TALLY	DC	0		
SUM,	0000			SUM	DC	0		
				BFFFF	DC	/FFFF		
				ONE	DC	1		
EXEC	1062 MICROSEC		2765 MICROSEC					
CORE	13 WORDS		20 WORDS					

FIGURE 3. SAMPLE PROGRAM CONVERSION PDP8 TO IBM 1800 ASSEMBLER

	LDX	1 99	SET COUNTER
	LD	DATA	GET FIRST VALUE
ADD	A	1 DATA	SUM 2,3,...100TH PTS
	MDX	1 - 1	DECREMENT COUNT
	MDX	ADD	LOOP
	STO	SUM	STORE RESULT
	WAIT		TARRY HERE

EXECUTION TIME 960 MICROSEC

CORE USED 7 WORDS

FIGURE 4. IBM 1800 PROGRAM WRITTEN FROM FLOWCHART

REFERENCES

1. Introduction to Programming PDP-8 Family Computers, The Software Writing Group Programming Department, Second Edition, (Digital Equipment Corporation, Maynard Mass., 1970).
2. PDP 11/20/15/R20 Processor Handbook, (Digital Equipment Corporation, Maynard Mass, 1971).
3. IBM 1800 Functional Characteristics, (IBM Systems Development Division, San Jose, California, 1970).
4. David Gries, Compiler Construction For Digital Computers, pp 412-433, (John Wiley and Sons, Inc., New York, 1971).

PLAN FOR NUCLEAR MEDICINE COMMUNICATIONS NETWORK

Francis K. Herbig, M.S.
Robert M. Donati, M.D.

St. Louis V.A. Hospital and
St. Louis University School of Medicine
St. Louis, Missouri

ABSTRACT:

A proposal to employ a communications system by which Nuclear Medicine data generated and stored at remote peripheral laboratories can be transmitted to a centralized analysis facility is presented. The need for this type of system is dictated by the geographical regionalization of health services and the lack of trained medical staff in remote areas. The system provides for transmission and analysis of proven clinical Nuclear Medicine procedures as well as the newer aspects of Nuclear Medicine, particularly dynamic imaging of regional time dependent concentrations of radioactivity. The hardware includes sufficient capability to accommodate new procedures in the foreseeable future without additional expense. The Nuclear Medicine procedures available and their interpretation will be comparable to those available in urban teaching centers. The basic technology to implement this program already exists; the program consists in joining communications, computer science and Nuclear Medicine technology to solve a specific medical problem.

PLAN FOR NUCLEAR MEDICINE COMMUNICATIONS NETWORK

INTRODUCTION:

Health care delivery in the United States is being critically reevaluated. As a consequence of the non-uniformity of the quality of health services many proposals have been developed. One plan, which has evolved, would provide for geographic regionalization of health services. Rutstein provides a succinct exposition of this concept. "Let me give you a simple outline of the plan. We shall emphasize the need for the regional structure of medical care programs, propose a more effective use of medical manpower, indicate ways in which the physicians' time may be more strategically applied, take advantage of rapidly advancing automation and technology, suggest added dimensions for medical research, and recommend a system of medical education to meet future needs. Not every hospital will be able to provide what every patient needs for the prevention and treatment of his disease. If this dilemma is to be resolved, we cannot build the medical system of the future around self-centered individual hospitals as we do today. Instead, through careful planning and efficient transportation and communication systems we have to weave together all medical facilities and personnel into regional medical service areas centered upon large teaching hospitals that are affiliated with medical schools."¹

The Veterans Administration Hospital system is in a unique position to test and evaluate this regional health concept. There is general consensus among those who have examined the present resources and long range objective of the nation's health care delivery system that the Veterans Administration must be constituted as an integral part of the solution to demand and supply in the health field. The basic goals and objectives, broadly outlined, for the VA hospital system include: improvement in the delivery of health services, improvement in the utilization of personnel in the treatment of patients, and in the operation of the health care systems through expansion of health services research, and extension of the VA's resources to community hospitals and physicians where the facilities and services do not now exist. To meet these needs in the context of an overall health care system the Veterans Administration has implemented a plan for regionalization of its health care services. This program calls for the establishment of core hospitals with strong professional specialty staffs, medical school affiliations, and ongoing research programs. These core hospitals will provide sophisticated technology and professional services to the non-university affiliated, primarily rural, Veterans Hospitals. The latter institutions would benefit from the integration with the core facility through ready availability of modern technology and professional medical consultation with resultant improvement in the quality of patient care. This organizational network can be extended through sharing arrangements to provide scarce health services to the community hospitals.

In accord with this concept, the Jefferson Barracks Veterans Administration Hospital and the John Cochran Veterans Administration Hospital, St. Louis, separated by 18 miles, have recently been consolidated. In assessing Nuclear

Medicine needs at Jefferson Barracks, it became readily apparent that a system which satisfied these needs might also meet the needs of the Veterans Administration Hospitals at Marion, Illinois and Poplar Bluff, Missouri. These hospitals do not presently, nor would they in the foreseeable future, have access to Nuclear Medicine facilities. The geographic relationship of these hospitals is shown in Figure 1.

FIG. 1 GEOGRAPHIC CONFIGURATIONS

MEDICAL TASKS:

To provide high quality Nuclear Medicine diagnostic services it is essential for a Nuclear Medicine specialist to interpret the patient data. Ideally, each hospital should have this type of physician. As a practical matter, there are insufficient numbers of Nuclear Medicine specialists attracted to non-urban areas. Therefore, we attempted to analyze the services which a physician would provide and to determine those aspects which could be accomplished by utilization of modern communications technology. Specifically, the area of greatest physician interaction is in the interpretation of radionuclide organ images and the maintenance of quality control in other Nuclear Medicine procedures. We would propose to use existing Nuclear Medicine and communications technology to acquire data at the remote facility and transmit these to the core medical facility for analysis and diagnostic interpretation.

In vivo, non-imaging procedures, such as the thyroid uptake, blood volumes, iron kinetics, etc., will be performed at the remote station by trained technologists following clearly defined laboratory protocols. Results, along with pertinent patient history, physical and associated laboratory data will be forwarded to the analysis station for critical review.

Samples for performance of in vitro assay procedures, thyroxine, B-12, angiotensin, thyroid stimulating hormone, etc., will be shipped daily via common carrier to the core hospital where a complete laboratory is dedicated to these tests. Centralization will provide comprehensive quality control and sufficient volume for ordinarily infrequently ordered tests to be readily available.

The major portion of the Nuclear Medicine communications system will be devoted to the acquisition and transmission of radionuclide images, both static and dynamic. The inclusion of dynamic capabilities for the peripheral laboratories increases the sophistication and expense of the network hardware. However, the peripheral station must provide, within the limitation of not having a Nuclear Medicine specialist in residence, capabilities as advanced as those available in medical centers. In addition, the system must be designed to accommodate new clinical techniques as they become available. Furthermore, a fully equipped remote Nuclear Medicine laboratory, which is easily converted to stand-alone capability, may make possible the recruitment of a Nuclear Medicine specialist in the future.

BASIC PRINCIPLES OF THE SYSTEM:

Prior to detailing the specific hardware and software, requirements needed to perform the medical tasks, some discussion of the general principles which the system should satisfy is in order. The system will link the core analysis station for processing and interpretation of data physically located at the John Cochran Division of the St. Louis Veterans Hospital with the Nuclear Medicine laboratories at the John Cochran, Jefferson Barracks, Marion and Poplar Bluff Veterans Hospitals (see Figure 1). The information flow will be from the station to the center with no data transmission between stations. Two way communications between the center and the stations is required, but the major flow of data in terms of volume will be from stations to the center.

The system must be extremely easy to operate; must not require the availability of specialized personnel for routine operations. The system must possess all the capabilities and features required to produce the desired results and to increase substantially the diagnostic capability of the physician. The time required by the specialized Nuclear Medicine physician to operate the system should be as small as possible. This specialist should be able to review a days procedures in no greater time than if they were performed at the laboratory of the central facility. Capability for future expansion and modification with no disruption of the functioning system should be an integral part of the system. The components should be available and well tested without need for extensive modification. The intent of this system is to join the knowledge of data processing, communications, and Nuclear Medicine technology to accomplish the goals, not to develop new technology.

COMPONENTS AND TASKS OF THE IMAGING SYSTEM:

The requirement for dynamic imaging capability has led to the selection of the gamma camera as the primary imaging device for the peripheral laboratories. The camera will be interfaced through an A.D.C. to acquisition hardware. The A.D.C. digitizes into a unique core memory address, the x and y position signals generated in the gamma camera. The contents of the location addressed is incremented by one each time it is addressed. The information thus obtained is stored as a square array of words containing count information corresponding to positions of scintillation events taking place in the camera crystal. The acquisition system will be capable of acquiring 64 x 64 matrices at a rate of at least five frames per second. A number of such camera data acquisition systems have been designed and are commercially available. In addition to the standard features found on such a system, an interactive display routine is required for the technologist prior to commencing the patient study. Patient identification,

study parameters, and relevant patient data to assist in the interpretation will be entered and tagged to the patient record. A standardized prompting routine will assure completeness and uniformity of data for each study. Following the termination of a frame of data on a preset time or count the image matrix is read onto a peripheral mass storage device, magnetic tape or disc. Peripheral laboratories configured in this way will include the clinical laboratory of the core hospital as well as those in outlying hospitals. A block diagram of the hardware configuration at the peripheral laboratory is presented in Figure 2.

FIG. 2 HARDWARE CONFIGURATION - PERIPHERAL LABORATORY

The image data collected at the remote laboratory sites will be transmitted to an analysis center located in the core hospital. Consideration was given to transmission via dedicated video-grade lines from the acquisition station to the analysis center. This would allow an initial reduction in peripheral hardware and expense and permit receipt of data at the analysis station essentially in real time. However, the high cost of modems, the continuing high expense of leased lines of this quality, and the fact that these lines do not exist in all areas to be served, led to the selection of voice grade communication lines as the medium for data transmission. Transmission at 2400 baud rates on unconditioned, dial-up phone networks is feasible with a number of available modems. The data processing hardware then must serve a dual purpose, acquisition and communications at the peripheral laboratory, and communications and display at the analysis center.

The communications package for the system, the modems, interfaces, and software, will generate text for synchronous transmission. This method was chosen over the somewhat simpler asynchronous mode since synchronous protocols generally call for transmission of large data blocks followed by a block check code which is small compared to the size of the data block. In contrast,

asynchronous protocols require start-stop elements for each word or character, with an overall reduction in transmission efficiency. The block check code is used to check data received against data transmitted. There are several methods of generating a code which reduce the undetected error rate below one bit in 10^7 with relatively small redundancy. Attention is given only to error detection. Upon detection of an error the processor will request retransmission of the data block. Although error correction codes are possible, the redundancy often approaches 50% of the transmitted data with a consequent reduction in transmission efficiency. The programs for the remote stations will specify files to be transmitted and initiate transfer following automatic answer, will log calls and time, and will restart following interrupted transmission which is likely with long connect times. The remote laboratory technician can gain emergency access to the analysis center for transmission of studies requiring immediate interpretation. A functional diagram of the software at the peripheral laboratory is presented in Figure 3.

FIG. 3 PERIPHERAL STATION SOFTWARE

The function of the analysis station is the receipt and display of the framed data transmitted from the remote laboratory terminals. The communications protocol for this station will nominally access all remote stations on the network separately in the evening and night hours via an automatic call unit and stored dial-up numbers. Data received will be error checked, acknowledged and filed by patient and hospital. Additionally, an index to the patient file listed by hospital will also be generated in order to permit rapid access to specific patient information. The system will also initiate a reconnect in the event of disconnect and abort the transmission if a high error rate, necessitating many retransmissions, is encountered. A log of such activities will serve to detect marginal lines and equipment. The equipment configuration for the analysis station is shown in Figure 4, and the software functions are indicated in Figure 5.

FIG. 4 HARDWARE CONFIGURATION - ANALYSIS STATION

FIG. 5 ANALYSIS STATION SOFTWARE

After receipt of the pertinent data, certain data processing routines such as flood correction, smoothing images, and selection of display contrast for static images can be effected under unattended operation. This will permit rapid photography and preparation of static images for interpretation as the first daily function of the analysis station. Dynamic studies will require the interaction of the Nuclear Medicine specialist and the display capabilities required are consistent with those now available on commercial systems. These include the ability to display the data frame by frame, addition or subtraction of frames, display of count versus time, information from a region of interest, and to display a count profile from a region of interest. Programs for quantitative analysis of dynamic studies can be added to the system. Reports to the remote laboratories will be prepared off-line, loaded and transmitted on a predetermined schedule.

HUMAN ENGINEERING:

One of the primary problems in the initiation of such a program will be the lack of qualified trained personnel in the areas to be served by such a communications network. We plan to obviate this difficulty by recruiting potential technicians from these geographical areas and training them in our ongoing Technician Training Program. Applicants who will be considered would be already employed at the peripheral location as x-ray technicians or medical technologists. If this plan is not feasible, similar individuals in the geographical area not associated with the specific peripheral hospital will be sought. The training program would be so designed as to produce technologists who upon completion of their training can assume the full responsibility for the technical operation of the Clinical Nuclear Medicine Laboratory in the peripheral facility. The technician will have gained the necessary knowledge to handle minor mechanical problems that might develop and to recognize possible radiation hazards and take the appropriate action. Training will be task-oriented requiring little decision making on the technologist's part. The hardware, software system will assist them in setting up all imaging procedures and there will be little margin for error. Limiting the training to specific instruments and a specific system will provide for economy of training time. It is estimated that a training period of 4 months duration would be sufficient. The training period will rely heavily on practical clinical experience and gaining technical proficiency in diagnostic procedures that the trainee will be required to perform at the network hospitals.

SUMMARY:

We have presented a proposal to employ a communications system by which Nuclear Medicine data generated and stored at remote peripheral laboratories can be transmitted to a centralized analysis facility. The need for this type of system is dictated by the geographical regionalization of health services and the lack of trained medical staff in remote areas. The system provides for transmission and analysis of proven clinical Nuclear Medicine procedures as well as the newer aspects of Nuclear Medicine, particularly dynamic imaging of regional time dependent concentrations of radioactivity. The hardware includes sufficient capability to accommodate new procedures in the foreseeable future without additional expense. The Nuclear Medicine procedures available and their interpretation will be comparable to those available in urban teaching centers. The basic technology to implement this program already exists; the program consists in joining communications, computer science and Nuclear Medicine technology to solve a specific medical problem.

REFERENCE:

1. David D. Rutstein, The Coming Revolution in Medicine, (M.I.T. Press, Cambridge, Massachusetts, 1967).

DATA ACQUISITION AND DISPLAY TECHNIQUES

Samuel Wilensky

Hybrid Systems Corporation
Burlington, Mass.

ABSTRACT

The digital-to-analog converter is discussed in regard to using the converter in a computer driven display system. Converter parameters, such as settling time, linearity and resolution, are discussed and examples are given as to how these parameters effect the oscilloscope display.

April 1972

DATA ACQUISITION AND DISPLAY TECHNIQUES

Samuel Wilensky

Hybrid Systems Corporation
Burlington, Mass.

Nearly every computer system involved in the processing of nuclear medical data has an analog output device for presenting data in a useable and easily understood format. The output device usually consists of three components: 1) the logic elements, usually part of the computer, which form the data into digital words; 2) the digital-to-analog converter (DAC) which transforms the digital data to an analog signal and 3) the display element such as an oscilloscope which is driven by the DAC. The DAC is the least understood of the three elements, so the following will try to expose some of the esoteric mystery which surrounds DAC's. The paper will discuss the DAC parameters which are of importance when specifying units for a display system.

In most display systems which are of a refresh variety, SETTLING TIME is a very important parameter. Settling time is defined as that duration of time required for all transients to have diminished to a sufficient degree so that the output is within some specified amount of the final value after the initiation of the change.

In most cases, the allowable deviation from the final value is generally the linearity which is $\pm 1/2$ LSB. However, in the case of DAC's with only a few bits or a converter with many bits, the settling time is usually specified as a per cent of full scale. A 12-bit D/A converter which is specified to settle to 0.1% in 200ns will settle to 8 LSB's of its final value in 200ns. Depending on the design of this DAC, it may fully settle to $1/2$ LSB in $1\mu s$ or it may never settle to $1/2$ LSB of 12 bits if the output is noisy. The settling time will be a function of the type of switches, resistor and output circuitry used.

The current mode switch offers the greatest speed potential. Wire wound resistors, though highly stable, tend to produce longer settling times because of inherent inductance. Film or deposited resistors are the choice if speed is of utmost importance. The DAC giving the best performance for display systems is a current output device with a terminating resistor and no output amplifier.

A typical high performance current output DAC should settle to 0.05% in 50ns and have an output current of 5 ma. Devices meeting these specifications are available from a number of different manufacturers.

In the discussion of settling time, the total time for the transients to settle out was considered, but not the nature of these transients. As much as we would like to think of DAC's having nice RC exponential (first-order) approaches or even with a little overshoot, to each subsequent output level, they can actually experience violent spikes, or "glitches," during the transition period.

Naturally, in our capacitive coupled world, we cannot escape some amount of feed-through of the rapid rise and fall times of the digital signals appearing at the output of the DAC. However, a greater source of disturbance exists which results from the fact that transistors turn-on and turn-off times are often unequal. This means that, for an instant after a new digital number is applied to a DAC, a temporary state can momentarily exist which sends the output, perhaps violently, temporarily in the wrong direction.

If, for example, the turn-on time exceeds the turn-off time, a "0" to "1" transition of a bit momentarily stays as a "0" (illustrated as 0 - o - 1) whereas a "1" to "0" transition goes directly to the "0" (illustrated as 1 - o - 0). Table 1 shows the intermediate state of a "o" which occurs during a transition from binary 3 to binary 4. Thus, instead of going smoothly from the output corresponding to 3 to that of 4 (one incremental step!), the DAC instantaneously swings towards zero before getting back to where it belongs..

.....	0 1 0 0 = 4	
.....	o o o o = 0 = intermediate state	
.....	0 0 1 1 = 3	
	digital	assuming:
	input	turning on turn off

Table 1. Intermediate transition states.

the time for this meandering being a function of the turn-on time. A little thought will convince the reader that the transition producing the worst temporary excursion is between the states:

100.....0
011.....1

In this case, the temporary state of all "0's" can cause a complete half-scale "glitch".

By utilizing rapid symmetrical switches, the width of these glitches can be reduced to less than 20 nanoseconds. If the output amplifier and subsequent circuitry are slew-rate or bandwidth limited, the effect of the glitch will be partially filtered out. Naturally, the sacrifice is in overall DAC response time, which may or may not present a problem in a specific application.

Conceptually, a track-hold would provide a "de-glitching" function by holding the previous value of the DAC until the glitch has died out and then acquiring and tracking the correct new state. The transient in the track-hold as it goes from the previous value to the new one can be made relatively smoothly. The control signals for the track-hold would be the leading and trailing edge of the clock pulse which strobes the new digital input into a buffer storage providing such exists in the system. Obviously, the overall settling time of the output track-hold. When designing a de-glitcher, one

must make sure that it does not, in reality, generate equally undesirable transients. Generally speaking, however, the "glitches" are sufficiently narrow or small and can be ignored in most applications or can be "blanked" by getting the oscilloscope off during the transition period.

The other parameter, besides settling time, which is of paramount importance in selecting a DAC for a display system is the precision of the unit. The DAC user, when trying to assess the ability of a particular DAC to convert a digital number to an analog voltage is faced with the not insignificant task of interpreting and evaluating the manufacturer's specifications. How accurately the DAC performs the conversion of a digital number to an analog signal is usually specified by some combination of three parameters. The parameters are resolution, linearity and absolute accuracy.

RESOLUTION

The term RESOLUTION is universally understood to refer to the number of discrete analog states which can be obtained at the output. For example, a DAC with 10 binary bits has 2^{10} or 1024 discrete output states. The resolution in this case might be referred to as "10 bits" or "0.0%", the latter derived from 1 part in 1024.

LINEARITY

As seen above, the term resolution is an unambiguous term which is usually well understood by both the manufacturer and user, but the definition of LINEARITY, on the other hand, is probably the least understood of any of the DAC specifications. The essence of the linearity specification is to describe how closely the DAC will approach an ideal straight line transfer function. The misuse of the term linearity usually arises because there are two kinds, integral linearity and differential linearity. To understand the difference between the two linearities, let's assume we have a 4 bit DAC with the following bit weightings:

	<u>Actual Weight</u> <u>(Volts)</u>	<u>Ideal Weight</u> <u>(Volts)</u>
Bit No. 4 (MSB)	8	8
Bit No. 3	4	4
Bit No. 2	1	2
Bit No. 1 (LSB)	1	1

If a 4 bit binary counter were used to drive the DAC as shown in Figure 1 the transfer function shown in Figure 2 would be obtained.

The line AB which is drawn so that it is equal distance from the upper and lower sets of points is a straight line which passes within 1/2 a vertical division of each point. Therefore, since the LSB has a weighing of 1 volt, we can say that the DAC has an integral linearity of $\pm 1/2$ LSB. That is to say, the maximum deviation between a fitted straight line and the transfer function of the DAC would be less than or equal to $\pm 1/2$ of the least

FIG. 1

FIG. 2

significant bit. It must be kept in mind that the best fitted straight line does not necessarily pass through zero or full scale; in fact, in the case illustrated the straight line passes through $-1/2$ volts at zero digital input and $14 \frac{1}{2}$ volts at full scale digital input.

Examining the transfer function, it is obvious that, even though a fitted straight line will pass within $1/2$ LSB of the transfer function, there are adjacent changes of state whose differences are either 2 LSB's or zero. These zero changes occur at digital changes 1-2; 5-6; 9-10; 13-14 and the 2 LSB output changes occur at digital changes of 3-4; 7-8; 11-12. We can say that this DAC has a differential linearity of ± 1 LSB* or that each unity change in the digital number will result in a change in the output voltage of an $\text{LSB} \pm 1$. The differential linearity in effect defines the local change in the slope or derivative of the DAC transfer function.

We can then see that the integral linearity defines how close a straight line can be fitted to the DAC transfer function, whereas the differential linearity is a measure of minimum and maximum size of each output step size. The differential linearity is usually twice as large as the integral linearity. Therefore, a DAC which has an integral linearity (that is usually what the linearity specifications refer to) of $\pm 1/2$ LSB could have a differential linearity of ± 1 LSB.

Along with linearity, one usually finds on a specification sheet that a DAC is monotonic which indicates that each successive output state is greater than the preceding one. Therefore, a DAC with an integral linearity of $\pm 1/2$ LSB or a differential linearity of ± 1 LSB is monotonic, but since monotonicity only refers to the minimum change, a monotonic DAC does not necessarily have an integral linearity of $\pm 1/2$ LSB. Therefore, it is redundant to specify monotonicity for a DAC with a linearity of $\pm 1/2$ LSB.

A DAC application which dramatically demonstrates the relationship of resolution and linearity is that of computer generated displays such as those used in remote consoles. In this application, a raster is generated by the computer appropriately stepping DAC's connected to the X and Y axes of a display console. These raster displays often contain as few 128 lines (i.e. 7 bits of resolution) but the spacing between the lines must be as nearly equal as possible or nonuniformity in the raster can easily be detected by the eye. This phenomenon is graphically demonstrated in Figure 3 which shows four 128 x 128 rasters generated by a computer display. The DAC generating the Y sweep has a linearity of $\pm 0.015\%$ (12 bits) and the DAC generating the X sweep has a linearity varying between $\pm 0.015\%$ (12 bits) Figure 3a to $\pm 0.4\%$ (7 bits) Figure 3d.

*This is not exactly correct since the full scale output is 14 volts and not 15 volts, which makes the LSB actually $14/15$ of a volt, but if one is inclined to be rigorous, we should also use the distance of closest approach of the line to each point and not the vertical distance which would more than compensate for the LSB of $14/15$ of a volt.

FIG. 3

It is obvious that a 7 bit DAC with $\pm .4\%$ linearity (7 bits) would not be adequate. In fact, a linearity of at least 0.0125% (12 bits) is necessary to generate a raster which looks uniform to the eye. Therefore, one should be very cautious when specifying DAC's which are to be used in raster type displays for the linearity must be several fold better than the number of bits of resolution if a uniform display is to be provided. In practice, this usually requires one to buy a DAC with more resolution than is necessary in order to satisfy the linearity requirements. For the case illustrated, it was necessary to use a 12 bit DAC, although only the 7 most significant bits were used.

ABSOLUTE ACCURACY

So far, in our discussions on the accuracy of DAC's, we have only considered how close the transfer function of the unit approximates a straight line. We have not concerned ourselves with the value of the slope of the straight line or how near it comes to passing through zero. The ABSOLUTE ACCURACY is the term used to define how close the analog output actually represents the digital inputs. Absolute accuracy includes, not only such errors as linearity, but also scale factor and offset. The scale factor and offset are governed by factors such as a zener voltage and amplifier offset voltages. For accuracies better than 0.1% , provision for external trimming should be provided, since zeners diodes tend to drift with time and power supplies are usually unbalanced enough to require external trimming.

The specification for absolute accuracy is usually given as a percent of full scale. Therefore, a unipolar DAC with a full scale output of 10 volts and an absolute accuracy of $.1\%$ could have an error as large as 10mV. Therefore, the voltage output for a zero digital code would be $0.000 \pm 0.010V$. If the $.1\%$ DAC had a bipolar output of +10 to -10 volts, the error could be as large as 20mV, since the accuracy is usually expressed as a percent of the full scale span (F.S.S.).

The user should be aware that, in most cases, the absolute accuracy is of secondary importance to the linearity, since in most systems there are gain and offset adjustments available, but the linearity is a property of the DAC itself, and is not usually available for adjustment.

**The Use of Video Systems for Acquisition
and Display of Static and Dynamic Function
IMAGES**

**A. B. Brill, P. Richardson, J. Erickson,
J. Patton, and J. Parker**

ABSTRACT

A video system for acquisition and display of data from nuclear medicine and radiographic studies has been developed at Vanderbilt University. The design of the system is presented along with examples which show the type of high resolution images that can be acquired and displayed with the system without a large computer core storage allocation.

**VANDERBILT UNIVERSITY
Nashville, Tennessee**

**Supported in part by AEC Contract AT-(40-1)-2401
and an University Cooperation Grant from NASA**

**The Use of Video Systems for Acquisition
and Display of Static and Dynamic Function
IMAGES**

**A. B. Brill, P. Richardson, J. Erickson
J. Patton and J. Parker**

The system requirements for on-line acquisition, analysis, and display at high resolution radiological and radionuclidic images can be met using currently available technology. The use of mini- or midi-computers along with different types of analog and/or digital buffers makes it possible to collect, process and display images collected at high data rates, as sequential or random events with array sizes in excess of 512 x 512.

In order to collect and analyze data from rapid transients measured with the Anger-type Scintillation Camera, and with image intensifiers in angiographic studies, we have acquired and interfaced a video disk and an electrical storage tube to a PDP-9 computer for data acquisition and display buffering. The configuration of the system and preliminary results with these systems are presented in this report.

The development of large array flicker-free displays requires some type of buffer for display refresh. This is needed in order to free the computer for other tasks, as well as to minimize the cost of the facility. In some cases, the functions of the display system can be enlarged to include data acquisition into the display buffer. The functions provided by our "display system" lie in the 3 areas listed below:

- (1) Data Acquisition
  - A. Monitor data collection
  - B. Buffer
- (2) Analysis
  - A. Program Development
  - B. Image Processing
- (3) Display Generation
  - A. Monitor
  - B. Hard Copy

During data acquisition in the clinic a monitor display of the computer or buffer stored data are needed. This should indicate that the study is being collected properly, and should provide information as to the findings so that study procedures can be modified appropriately. During the data analysis phase, a higher quality display is needed than for monitoring purposes. While developing programs, and evaluating different algorithms, it is necessary to generate high quality images and to compare those to the unprocessed data as collected by the imaging device. In the past,

computer-generated image displays for nuclear medicine were frequently of lower quality than the usual analog displays. This no longer need be the case. During image processing, one has to know when to utilize different options in the analysis package and a high quality monitor of the image is needed to guide processing. This same function is shown as the monitor function during display generation when routine processing is used to produce a picture for the patient's record.

The different types of display systems used in Nuclear Medicine at Vanderbilt University are shown in Table I. Three different types of buffer memory devices are in use:

- (1) Intertechnique 4096 Multichannel Analyzer (BM-96): Coupled to PDP-9.
- (2) Intertechnique (BM 96) - Tolana 15 track Instrumentation Grade Tape Recorder.
- (3) Intertechnique (Model 33) Cinescintigraphy system.

Each of these 3 separate systems can collect static or dynamic function studies, and permits interactive display manipulation, such as the generation of time rate of change curves. The systems permit the storage of dynamic function data for 64 x 64 resolution arrays for >30 sec/frame, >10 msec/frame, or >250 msec/frame respectively. Direct access into the computer from the camera ADC permits the same capabilities as does the second system listed although direct entry into the computer incurs less dead time per pulse.

The data in the computer are displayed on a Tektronix 503 oscilloscope with a P-11 phosphor. Flicker free display of small sized arrays (i.e. 64 x 64) are obtained with this system. Higher resolution images are written on the oscilloscope and viewed by photographic integration. The line printer provides surprisingly high quality intensity coded images rapidly and is the best display monitor we have achieved to date. The use of calcomp plots has been helpful for the generation of high quality line drawings for publications. Color has been employed with success for time coding dynamic function studies, and warrants further investigation.

The video buffering systems are the subject of the present report. The video disk we are using is Data Disc 3104V, 1800 rpm, 4 MHz bandwidth (frequency at which the recorder gain has decreased by 3dB). It can buffer store 600 separate TV frames. The data from two separate cameras can be collected, each at a maximum of 1/30 frame rate. A maximum of 600 frames can be collected, which corresponds to 20 seconds of real time data acquisition at 30/sec frame rate. With this system, data are collected from image intensifier cameras in the cardiac catheter lab and by TV cameras viewing the Anger Camera CRT. An interface has been designed and built to digitize an entire frame at 256 x 485 resolution in approx 4 seconds. Curves can be set with thumbwheel switches to digitize the activity

DISPLAY SYSTEMS USED IN NUCLEAR MEDICINE AT VANDERBILT

(1) Buffer Memory with Interactive Contour-Volumetric Display

- A. Monitor/Data Acquisition
- B. Time rate of change curves

(2) Computer - CRT

- A. Monitor - (32 x 32 x 6)
- B. Photogeneration - to 1024 x 1024 x 64
- C. Line Printer
- D. Calcomb Plotter
- E. Color (spatio - temporal)

(3) Video Buffering Systems

A. Video Disc

(1) Data Acquisition

- a. Roentgen Videometry
- b. Nuclear Medicine image intensifier studies

(2) Data Display

- a. Static images
- b. Dynamic images. (stop frame motion pictures)

B. Electrical Storage Tube

(1) Data Acquisition

- a. Scan Raster
- b. Random Access

(2) Data Buffer For Display

- a. Single image: <1024 x 1024 x 10
- b. Multiple images: Static/Dynamic

256x256x16
512x512x4
128x128x64

Fig. 1. Schematic diagram of Vanderbilt Nuclear Medicine computer configuration emphasizing the video interfaces.

distribution in any rectangular subregion of the image.

Figure 2 is a block diagram which shows how a rectilinear scanner or Anger Camera could be coupled directly to the EST. At present, the data would then have to be transferred to the video disc for digitization (4 sec) and transfer into the computer.

Two studies will be presented which show the capabilities of the system. Figure 3 is a 256 x 256 image of a thyroid phantom. Data were collected into the computer from a rectilinear scanner, transferred to the EST, displayed on a TV monitor, and time exposure photographed with a Polaroid camera. The image was of superior quality as viewed on the monitor. Only the central portion of the EST was used in this image.

An Anger camera study of a heart bolus was collected on digital magnetic tape. Sixty seconds of data were stored at 250 msec/frame, in a 64 x 64 array. Sixteen frames are displayed in Figure 4 from this study. Each represent the normalized data from six summed frames (1.5 seconds). By normalization it is possible to obtain a good view of all frames at a single photographic intensity setting. Each frame is displayed as an 128 x 128 image, and the central 512 x 512 portion of the EST was used. If additional frames were desired, a 4 fold increase in display frames could have been achieved with the same size matrix elements.

With the ability to record and display composite images, it is clear that multiple views of a single organ could be displayed simultaneously. Computer processed vs raw data could be viewed. Along with images, text can be written with the scene, either by computer character generators or by the use of special effects generators.

The system as it exists in our laboratory is in an early phase of development. No serious problems have yet been identified which compromise the potential of the system, although, it has not yet been tested in all modes. Costs of the electrical storage tubes are in the \$500-\$1000 range, and packaged as systems, they cost upwards of \$4000(PEP). Costs will undoubtedly fall as more experience is obtained. The video disk we are using cost \$32000 (DATA DISC). This is considerably higher than can be justified for a single application. Costs have been reduced markedly in the last year, and it is likely that systems will be available at less than \$10,000 in the near future. These systems bear careful consideration in the design of data acquisition and display systems for use in Nuclear Medicine.

ACKNOWLEDGEMENT:

Supported in part by AEC Contract AT-(40-1)-2401 and an University Cooperation Grant from NASA

Fig. 2. Schematic diagram of proposed use of an Electrical Storage Tube (EST) for Nuclear Medicine image collection and display.

Fig. 3. Image of Picker Thyroid phantom collected in the PDP/9 in a 256 x 256 array. The image was written onto the EST, and displayed flicker-free on a TV monitor and photographed with polaroid film.

Fig. 4. Multiple images of heart bolus study. Each 128 x 128 image represents a 1 1/2 sec time interval. The intensity was normalized to 64 counts in the maximum cell, and displayed as in Fig. 3.

GAMMA-CAMERA-TO-COMPUTER INTERFACING -- THE EASY WAY
T. R. Barclay, A. C. Morris, Jr., A. E. McDow, Jr.
Medical Division, Oak Ridge Associated Universities, Inc.
Oak Ridge, Tenn. 37830

ABSTRACT

A relatively inexpensive and fast acting electronic interface for linking a scintillation camera to a digital computer has been developed. Gamma rays radiating from a patient produce signals in the Pho-Gamma III detection electronics. Processed signals are transmitted to an interface module located near the computer. The interface electronics inhibits the gamma camera electronics while analog-to-digital conversion of X and Y coordinate signals for a gamma event is accomplished: resulting digital signals are routed on to the IBM 1800 computer. The stored image of the scan can be displayed on a TV monitor or in printed form. We believe this camera-to-computer interface is economical, easy to reproduce, and adaptable to many types of digital computer systems, both to large systems and to minicomputers of the PDP variety.

GAMMA-CAMERA-TO-COMPUTER INTERFACING -- THE EASY WAY

T. R. Barclay, A. C. Morris, Jr., A. E. McDow, Jr.

Medical Division, Oak Ridge Associated Universities, Inc.

Oak Ridge, Tenn. 37830

The goal of instrumentation for diagnostic clinical scanning is to produce high-resolution images showing the distribution of radionuclides in various organs. As detection and analytical instrumentation become more sophisticated a need for more precise readout is realized. Many manufacturers now offer scanning instrument systems that include digital computers.¹⁻³ Through use of the large storage capability of these computers it is possible to obtain rapid reproduction of high resolution images from scanned areas.⁴ The state of the art in scanning techniques and computer applications is rapidly changing, so many institutions with computers do not consider it economically feasible to acquire separate computer equipment for scanning purposes. Instead, such institutions have developed interface electronics linking scanning equipment to existing computers.

The gamma-camera interfacing technique described here is part of a continuing effort at our facility to connect all clinical scanning instruments to our IBM 1800 digital computer.

CONSTRUCTION AND OPERATION

Our Pho-Gamma III gamma camera was modified to permit remote control of the pulse-height-analyzer and scaler/timer circuits. We obtained analog output pulses from the X- and Y-axis deflection and unblanking pulse circuits of the system's oscilloscope. The X and Y signals are bipolar and vary in amplitude ± 2.5 volts to give the coordinate position of each detected event. An unblanking pulse, which is synchronized with the X and Y signals, normally gates the observation oscilloscope on for a few microseconds. The location of the light flash on the face of the oscilloscope cathode-ray tube corresponds to the position of the detected event in the camera's crystal. A parallel connection of these signals is routed to an interface module located 100 ft away at the computer site; here the signals are converted to digital form for the IBM 1800 input terminals.

Figure 1 shows the data and control signals required for operation. With a patient positioned under the camera head, the operator places a Local/Remote switch on the camera control panel to Remote. The Service Request pushbutton is pressed to ready the computer's appropriate data acquisition program. This accomplished, the computer signals by turning on a Computer Ready light, and the operator can initiate the picture accumulation by depressing a Start pushbutton. When the computer receives the Start signal a computer-generated signal resets all the interface control flip-flops, which also removes the Inhibit signal from the pulse-height-analyzer of the Pho-Gamma camera. Gamma events are then processed; the X and Y deflection signals for each event are routed to the analog-to-digital converter circuits

1. Pho-Gamma III to IBM 1800 computer interface signal flow and block diagram.

of the interface. At 1.5 microseconds after receipt of the data pulses, they have been sampled and an Inhibit signal is generated in the interface to stop processing subsequent events until the interface is again ready. During these few microseconds, the interface completes its analog-to-digital conversion, calls the computer to receive the data, and uninhibits the camera to process the next gamma event. The entire operation takes 27 microseconds per detected event. A picture accumulation is terminated when the operator presses the Stop pushbutton, or when a preset number of events are analyzed.

The interface equipment consists of an interface rack, plug-in boards, and power supplies. Rack, power supplies, and mounting hardware cost \$688. Components for special purpose boards were \$8.50, and standard electronic plug-in boards were purchased from the Digital Equipment Corporation for \$1,677. This price included sets of Sample and Hold circuits, analog-to-digital converters; and other one-shot, flip-flop, NOR-Gate, and Driver boards required for the system (Fig. 2). Cost of modifying the Pho-Gamma and necessary interconnecting and control cables was \$200; giving a total interface hardware expense of \$2,553.50.

Connecting the various interface circuits was accomplished using the wire-wrapping process. Such an operation would be easy for an electronics technician to perform in a week or less, because the time-consuming portion of the project (engineering) has already been completed by the ORAU staff.*

PROGRAMMING AND RESULTS

The computer is programmed to accumulate and store digital data from each detected event, and to provide a high-speed printout of the radioactive image using alphanumeric symbols. Display of the data is also accomplished through use of a Tektronix 4501 scan-converter and remote TV monitor. Figure 3 shows examples of (a) oscilloscope readout of the Pho-Gamma III camera, (b) computer-produced image taken from the Tektronix 4501 scan-converter, and (c) a computer-assisted enlargement of the image (b).

CONCLUSIONS

We believe this camera-to-computer interface is economical, easy to reproduce, and adaptable to many types of digital computer systems, both to large systems and to minicomputers of the PDP variety.

---

*Detailed diagrams of board interconnection and other circuits are available from the ORAU Medical Division on request.

2. Completed interface rack, power supplies, and plug-in boards.

3. (a) Pho-Gamma III image of ^{131}I phantom taken from camera's oscilloscope.
- (b) Computer-produced scan image of (a) taken from screen of a Tektronix 4501 scan-converter.
- (c) Computer enlargement of the thyroid image in (b).

REFERENCES

1. S. M. Pizer and H. G. Vetter. "Processing Radioisotope Scans," J. Nucl. Med. 10, 150-154 (1969).
2. W. N. Tauxe. "Digitization and Data Processing of Scintiscan Matrices by High-Speed Computer, in Medical Radioisotope Scintigraphy," Vol. 1, pp. 803-813 (IAEA, Vienna, 1969).
3. C. B. Cykowski, D. L. Kirch, C. E. Polhemus, and D. W. Brown. "Image Enhancement of Radionuclide Scans by Optical Spatial Filtering," J. Nucl. Med. 12, 85-87 (1971).
4. T. R. Overton, P. G. Heslip, P. A. Barrow, and J. Jelinek. "Dual Radioisotope Techniques and Digital Image-Subtraction Methods in Pancreas Visualization," J. Nucl. Med. 12, 493-498 (1971).

FAST DATA ACQUISITION WITH A CAMERA-COMPUTER-SYSTEM

F. Bitter, W. E. Adam

**University of Ulm
Ulm, Germany**

April, 1971

FAST DATA ACQUISITION WITH A CAMERA-COMPUTER-SYSTEM

F. Bitter, W. E. Adam
University of Ulm
Ulm, Germany

ABSTRACT

A survey on the requirements to a camera-computer-system will be given. The main parts with the most important features are described and the necessary data manipulation programs. The setup of the Nuclearmedicine in Ulm is outlined.

I. Data Acquisition

A camera-computer-system has to fulfill extremely different requirements. The spectrum of applications reaches from static investigations like brainscan to very fast functional studies. (Fig. 1). These applications may be classified into three groups.

- 1.) Dynamic studies (fast, slow)
- 2.) 2-dimensional averaging (direct, indirect)
- 3.) Static studies

1. In general fast dynamic studies are short, but last at least 10 seconds and up to 60 seconds. The time steps - the measurement-time per frame (picture) - are determined by the transfer rate and by the frame size, which will not exceed 4 K; that means a matrix of 64×64 points. The transfer rate should be about 40,000 words per second. These conditions lead to timesteps of 40 milliseconds for a 32×32 matrix and 125 milliseconds for a 64×64 matrix taking into account the maximum access time. Such fast applications require a core memory of double frame size and a peripheral mass storage of about 300 K 8-bit-words. Disk memory is suitable for it as well as tape memory.

In slow dynamic studies the transfer rate of the peripheral storage is not the critical point, but the capacity may have to be still greater. A liver study of 100 minutes, 20 seconds time steps and frame size 64×64 would take more than one million words of storage.

2. The second group - 2 dimensional averaging - has to be divided into two subgroups with different storage methods. First, you build up several frames in the core memory and the averaging process takes place by storing the data phase-coincident. With that periodic functions can be recorded. There are no peripheral storage requirements, but the core memory has to be as large as frame size times the number of frames, for instance

16 (32×32) - frames require 16 K of core memory
or
8 (64×64) - frames require 32 K of core memory

INVESTIGATION	TIME	TIME STEPS	FRAME SIZE	STORAGE		WORDLENGTH [BIT]	PULSRATE [PPS]
				CORE MEMORY [WORDS]	PERIPHERAL [WORDS]		
FAST DYNAMIC STUDIES (HEART)	10 SEC 10 SEC	40 MSEC 125 MSEC	32 x 32 64 x 64	2 K 8 K	250 K 320 K	8 8	100 K
SLOW DYNAMIC STUDIES (LEVER)	100 MIN 100 MIN	20 SEC 20 SEC	32 x 32 64 x 64	1 K 4 K	300 K 1 200 K	8 8	
STATIC STUDIES (BRAIN)	DEPENDING ON THE DESIRED QUALITY		64 x 64 64 x 128 128 x 128	4 K 8 K 16 K	< 100 K < 100 K < 100 K	12 12 12	
DIRECT 2-DIM. AVERAGING		≈ 30 MSEC	32 x 32	16 K	< 100 K	12	
INDIRECT 2-DIM. AVERAGING	DEPENDING ON PERIPHERAL STORAGE AND PULSRATE	ANY TIME STEP	32 x 32 64 x 64 64 x 128 128 x 128	< 4 K		10 12 13 14	< 10 K < 10 K < 10 K < 10 K

Fig. 1

This method may be used to obtain phase pictures of the heart. With the second method each pulse coming from the camera is digitized and its coordinates are stored away. In addition timing marks are stored. According to the desired resolution 10 to 16 bits storage are necessary and the pulse rate must not exceed 10,000 pps. Therefore the peripheral storage depends on the measurement time and the pulse rate; for 10,000 pps and 100 second measurement time one needs 1 million words. In this way the original information is stored and may be processed later on in any way.

3. Static studies place the least demands on the system. Little peripheral storage, generally less than 100 K words and a word length of 12 bits are sufficient. The core memory size is determined by the resolution, thus 16 K for a 128 x 128 matrix. Measurement times can be chosen almost arbitrarily. Figure 2 shows schematically a camera-computer-system.

Camera

Usually the scintillation camera delivers its information with two analog pulses - the coordinates (x,y) - and a digital signal (Z). For each γ -quantum of the right energy the camera generates these three signals. The pulse rate goes up to 300,000 pps concerning the whole spectrum and up to 100,000 pps within the photopeak.

ADC

Amplifiers adapt the analog coordinate pulses to both the analog and digital converters; the digital pulse starts them. Almost all ADC's on the market are useful and I want to point out only a few things.

- 1.) 8-bits resolution is sufficient
- 2.) The conversion time should correspond to the time resolution of the camera which is about 3 microseconds
- 3.) The linearity of all ADC's is good and so we have to look more to the differential linearity, which is not sufficient in ADC's that work after the principle of successive approximation. The counter-types are suitable, but they must work with 100 megacycles per second due to the required conversion time.
- 4.) A logical signal should show the end of conversion.

Interface

The real interface consists of gates, a buffer and a controller, which controls the transfer to the computer. With the gates you can alter the frame size and you get the possibility to magnify a sector of the picture. The final value, an address, will be stored in the buffer for synchronisation with the computer.

Computer

A computer of small or medium scale is suitable, which offers a direct memory access (DMA) channel and the possibility of "add 1 to memory" or "increment memory buffer" and the required memory size. If there is only one DMA-channel, a multiplexer is required, too, since the peripheral storage device uses the same channel.

Fig. 2

Peripheral storage

As peripheral storage you can use synchronous tapes with about 40,000 bytes per second (bps), disks and diskpacks with 50,000 words per second transfer rate. The capacity of these storage devices reaches from 250 K words for disks with fixed heads up to 20,000 K bytes (8 bits) for tapes.

Configuration of Ulm

The Nuclear-Medicine in Ulm uses the following equipment: As scintillation camera we have the Pho/Gamme III from Nuclear Chicago. 10-bit ADC's from Borer Electronics are adapted to the 1600 output of the camera and work with 100 MHz. A self-made interface connects the ADC's to the DMA-multiplexer of the computer, a PDP-12 from DEC with 16 K 12-bit words of core memory, 2 LINC tapes with 260 K capacity each, a point-display 9 x 9 bits and a real time clock. Within a few weeks this system will be expanded by a synchronous tape from PEC with a transfer rate of 36 K bps.

With this system it is possible to fulfill almost all requirements for data acquisition in the field of the Nuclear-Medicine.

II. Data processing

The data acquisition belongs mostly to the hardware but a suitable processing of these large data becomes a software problem, assuming that the computer is large enough to execute all the jobs. Comprehensive programs, such as are necessary for contrast enhancement and filtering, can hardly be done on small computers because the programming work will become a problem.

Now, let me continue with useful programs for processing functional studies by means of small computers.

Display

First of all the acquired data have to be presented on a display in various forms. A contour display with variable thresholds is very useful for selection of areas of interest. If there is enough information, the isometric display may give you a good survey. Slices through the pictures are advantageous, too.

Image processing

In functional studies you find very poor information per frame, so it is necessary to have the possibility of adding several frames and smoothing to equalize a little the bad counting statistics.

Figure 3 may give you an idea of how important it is to correct for detector inhomogeneities. On the left side you find two curves coming from different areas of a homogeneous distribution of activity. The two upper curves obtained from original data show two different areas and the third the difference between the curves. Below you see the same curves after smoothing and correcting the original data for detector inhomogeneities. Now the curves are almost equal. There is another example on the right side. The mean difference between the curves is about 30% uncorrected and about 4% corrected.

Usually, the differences are not as great as they are demonstrated here.

Fig. 3

Areas of interest

The programs we use allow us to obtain curves from 15 not overlapping areas at a time with or without smoothing and correction.

Processing of curves

Furthermore there are required programs for manipulation and comparison of these curves. They should allow smoothing, adding, subtracting, differentiating, integrating, extrapolating, calculating regression lines and, obviously, calculating transit times.

Output

A last group of programs is required for output of the reduced data. The simplest way is to photograph the data from display. Processed pictures can be written as symbolic scintigrams on a teletype, or, if available, with a lineprinter or plotter. With an XY-Recorder, we write curves.

DERIVATION AND COMPUTER DISPLAY OF REGIONAL
FUNCTION SCANS OF THE LUNG WITH $^{133}\text{XENON}$ *

William J. MacIntyre and Scott R. Inkley

Department of Medicine, Case Western Reserve University
and University Hospitals, Cleveland, Ohio

ABSTRACT

In the present study the relative contribution of each region of the lung to total effective gas exchange has been derived by combining the relative perfusion in the lung of intravenously injected $^{133}\text{Xenon}$ with the clearance rate of the Xenon from each individual site. Since effective gas exchange function is dependent both on the blood supply to a specific area as well as the evacuation of the gas transferred by the blood supply, this combined parameter represents the relative fraction of total lung function occurring at each of the sites measured. These values have been calculated for the entire lung field and displayed both as a 40×40 matrix as well as a three-dimensional plot with the z-axis representing the variation of the gas exchange parameter for each of the 1600 regions. This display of the product of the two variables has permitted an easy and accurate delineation and assessment of poorly functioning regions of the lung, while examination of the individual matrices can determine whether the reduced function is due to an abnormal perfusion or to a slow clearance.

*This material was presented at the Fourth European Course on Recent Progress in the Application of Nuclear Techniques in Biomedical Research, January 24-28, 1972 at Schloss Reisensburg, Germany

This investigation has been supported by the National Heart and Lung Institute Grant HE-06304, National Institutes of Health, United States Public Health Service and by the Mrs. Chester D. Tripp Foundation.

In conventional measurements of organ function or organ blood flow, external probes are placed so that their fields of view will encompass the entire organ. Following this placement, a single recording such as a heart dilution curve, a renogram, or a hepatogram, can be analyzed to reflect the total blood flow or function of that organ. Recently, various investigators have recognized the applicability of stationary detector devices, or scintillation cameras, to serve as multiprobe detectors. Thus, multiple heart dilution curves (1,2), renograms (3,4), and lung washout curves (5,6) have been recorded simultaneously by these systems.

The present report is concerned with utilizing multiple detection site capacity of the camera to demonstrate regional lung function. As an index of lung function we have selected the product of the amount of gas present at each site multiplied by the turnover rate of the gas at that site. As pointed out by DOLLERY et al in 1962 (7) the washout of Xenon following intravenous injection of the dissolved gas provides information primarily concerning the ventilation of perfused alveoli. Since the slope of the washout curve represents the fraction of Xenon cleared per unit time, the product of this slope multiplied by the amount of gas present in the alveoli by perfusion would then represent the total amount of gas cleared per unit time. By the same rationale the relative contribution of each region of the lung to total gas exchange may be calculated by using the characteristic slope of each element and multiplying by the fraction of total dose perfusing that element.

In addition, by means of multi-site calculation and display of these lumped parameters by computer, a complete representation of the entire lung field has been produced so that the clinician can visualize the relative contribution of each site of the lung to total function or gas exchange.

The methodology described above has been called functional scanning (8) or functional imaging (9) and exhibits two primary criteria.

1. The first requisite is that an integrated description of the entire organ must be obtained. This differs from the multiprobe technique in that the function of the whole organ should now be visualized by one representation, similar to a conventional scan display, and not dependent on correlating multiple curves or other parameters with sites from which the curves are derived selected by a separate visualization.
2. Secondly, the variation between adjacent sites must reflect differences in organ function. This means that the relative magnitude of each spatial segment must represent some accepted parameter such as a disappearance constant, gas exchange, clearance rate, or half-time that will reflect the actual function of the specific organ, not merely the amount of radioactivity deposited there.

METHOD

The distribution of $^{133}\text{Xenon}$ in the lung was detected by a

Nuclear-Chicago Pho Gamma II scintillation camera (Fig. 1) and accumulated by an RIDL 1600 channel analyzer in the form of a 40x40 word matrix (10).

Subjects were studied in both upright and supine positions after a rest period of at least 10 minutes in each position. They were positioned with back to camera on a comfortable stool and in the supine position on a canvas cot so that the bases were not elevated above the apices by the buttocks. The top of the camera collimator was positioned just below the spinous process of the 7th cervical vertebra. Injections were made through a plastic catheter placed in the left antecubital vein directed toward the superior vena cava and 3 mc. of ^{133}Xe in saline injected rapidly, followed by a 5 ml. saline flush. Following injection, the subject held his breath at resting inspiration (functional residual capacity plus tidal volume) with open glottis for a period of 10 to 15 seconds until a ratemeter monitoring the camera output indicated a plateau of radioactivity. The subject then resumed quiet, normal breathing while clearing the lungs of ^{133}Xe through an exhaust hood. Matrices of the distribution of ^{133}Xe in the lung were recorded every 2.8 seconds until the recorded amount was less than 20% of the peak value.

A conventional 1000 hole, low energy collimator was used for this study rather than the divergent collimator. For this reason, it is recognized that all portions of the lung may not fall within the field of view. A representation of a computer print-out of a normal lung together with a 10% contour line, however, shows that in most cases regions of the lung exhibiting an uptake of at least 10% of the peak counting rate are almost entirely recorded (11). This collimator has a diameter of about 26.5 cm. and by using a modification of the technique of OBERHAUSEN *et al* (12), the useful diameter of the crystal was equivalent. Each element of the 40x40 index thus corresponded to a square of sides approximately 6.6 mm. in width.

Calculation of Regional Function

To calculate regional gas exchange function, the perfusion of each region and the clearance of each region have been obtained by analysis of the washout curve wherein it has been found that the initial 60% of the curve can be usually fit by the single exponential expression:

$$Q = Q_0 e^{-\lambda t} \dots\dots\dots (1)$$

where Q_0 is the extrapolated counting rate at the time of the final frame of the plateau and λ is the clearance constant.

To obtain both of these parameters at every site, each element of the 40x40 matrices is initially subjected to a nine element smoothing routine involving bounding and averaging of the original counts in the matrix elements (13,14). The logarithm of the counts collected at each element, Q_i , on each frame is then fit by the method of least squares so as to obtain the optimum value of the disappearance constant, λ_i , for each element by the equation:

Figure 1. Schema for recording time-sequential distributions of regional radioisotope deposition in the lung following perfusion of $^{133}\text{Xenon}$.

$$\lambda_i = \frac{\sum_t t \log Q_i - \frac{1}{N} \sum \log Q_i \sum t}{\sum_t t^2 - \frac{1}{N} \left[\sum t \right]^2}$$

where $\log Q_i$ is the logarithm of the counts in each element of the matrix, t is the time at which the matrix was collected, and N is the number of matrices used for calculation of the slope. N will vary dependent upon the half time of washout but it has been found that eight matrices are usually sufficient. If the summation of the counting rate of the entire lung area represents total perfusion, the ratio of the counting rate from any site, Q_{oi} , to the total counting rate is the fraction of the total dose perfusing that area. The value of the extrapolated counting rate, Q_{oi} , is computed from the equation:

$$\log Q_{oi} = \frac{1}{N} \sum_t \log Q_i - \frac{\lambda}{N} \sum_t t \dots\dots\dots(3)$$

(Similar results could be obtained by collecting counts during the plateau period. Because of the low number of counts in each element, however, somewhat better statistics were obtained from the extrapolated values which were derived by least squares fitting of the clearance curve. In addition, the latter technique accomodates for the clinical subjects unable to hold their breath for a plateau reading.) In this investigation values of relative perfusion and values of relative clearance rates were obtained from each of the 1600 addresses in the 40x40 matrix. To obtain relative function of each site of the lung then required merely the multiplication of the two matrices, the first representing the relative amount of gas present and the second representing the rate at which the gas was exchanged.

Results

Relative values of 1600 element matrices for perfusion and the disappearance constant are shown in Fig. 2 normalized to a maximum of 20 (5% difference between levels). In the left figure, 2A, the large arrow designates a region that is clearing the ¹³³Xenon very slowly. The figure on the right, (2B) shows that adequate blood supply is demonstrated in the apex of the left lung in the same region that showed poor exchange of clearance.

Although the numerical matrices contain all the basic information of perfusion and clearance, quantification of the gas exchange function cannot be obtained until these two matrices are multiplied and the product visualized. Each element of this product scan thus represents the contribution to total lung function from that individual site, since it is derived from knowledge of the amount of gas deposited at that site (perfusion) and the turnover rate of the gas (ventilation).

This sequence is illustrated by studies on a patient with cystic fibrosis shown in Fig. 3. The lower level illustrates the perfusion matrix on the left and three dimensional model of this matrix on the right. Impaired perfusion of the left lung is apparent, as well as decreased ventilation as denoted in the center matrix and model. The effect on gas exchange is depicted in the top matrix and model which are the product of perfusion

Figure 2 A. Matrix of the disappearance constant, λ , recorded from a patient injected intravenously with ^{133}Xe on. Highest value normalized to 20 represents most rapid clearance. Arrow in left apex indicates record of poor clearance.

923214

20 SEC HOLD AT FHC

Figure 3. Matrix and corresponding three-dimensional representation of derived initial perfusion values (lower figure), the slopes of the clearance curve (middle figure), and the product of the two representing gas exchange potential (top figure). Impaired function due to cystic fibrosis.

and clearance and show that the left lung contributes very little to the total gas exchange.

The reduction in function cannot be immediately recognized from the slope matrix since it must be kept in mind that slopes are independent of the amount present and thus are plotted for all regions whether representing a large volume of active lung tissue or merely random background. The importance of clearance is directly related to the amount of gas present.

Similar results are seen in a measurement of another cystic fibrosis patient as shown in Fig. 4. Perfusion (the lowest level) appears reasonably normal but when multiplied by the values of the clearance slopes shown in the middle section, the gas exchange function of the right lung as shown in the top level, is considerably impaired. In addition, loss of function is also noted in the mid-lung region of the left lung (upper model), a fact not fully appreciated by the examination of individual representations.

Information of this type may be different to derive from static ventilation indices in the diseased lung when obtained by inhalation of $^{133}\text{Xenon}$ from a spirometer. The static ventilation distribution of the subject in Fig. 4 is shown in Fig. 5 as determined from a three breath inhalation at functional residual capacity (FRC) (15). The distribution appears reasonably uniform, demonstrating that distribution of inhaled gas is not necessarily alveolar and thus may not reflect deficiencies of gas exchange.

The dynamic ventilation scans are subject to a greater degree of random variation than the inhalation scan. The slope scans are quite reproducible, however, as shown by a comparison of slopes measured by clearance procedures on the same subject performed 100 days apart following resection of a right upper lobe bulla. The non-homogeneities of clearance in the lung are thus not due to random fluctuation but represent actual physiological differences.

Discussion

The unique information displayed in a single representation of perfusion of the lung combined with dynamic ventilation has brought an increased clinical importance to the established techniques of $^{133}\text{Xenon}$ investigations. This improvement has been ascribed partially to using the dynamic ventilation measurement (washout curves) which has a distinct advantage in the clinically involved patient as opposed to the static inhalation technique of ventilation assessment. A second advantage of the method is the three-dimensional display which allows the clinician to interpret directly the variation of gas exchange function in multiple regions of the lung. The selection of the three-dimensional scan as a readout is not, necessarily, a requisite of the functional concept. Other types of readouts may well be used, provided several criteria are met.

(1) The primary requisite is one of linearity. It is necessary that a wide range of values be visualized with equal perception, not merely that the presence of "hot" areas or "cold" areas be identified on a distribution scan. Accentuation

Figure 4. Three-dimensional representation of initial perfusion, clearance, and gas exchange values in second subject with cystic fibrosis. View of lung taken from opposite side of previous figure.

1619 16

20 SEC HOLD AT FRC

Figure 5. Normal appearing inhalation scan for patient illustrated in Fig. 4.

Figure 6. Comparison of slopes (disappearance constants) recorded on same patient before and after undergoing a surgical procedure for removal of a bulla in upper right section of the lung. Note the similarity of clearance in left lung regions in both determinations.

techniques, such as photoscanning, are difficult therefore, to adapt to this condition.

(2) Secondly, it is desirable to be able to ascertain small differences between values. Except in gross situation, function may not be expected to vary greatly between adjacent areas. Thus, it is important to be able to perceive small differences and to have the differences readily apparent to the eye. The numerical readout of the matrix is linear, but it is difficult to visualize readily small variations by reading numbers.

It is possible to obtain completely satisfactory function curves from direct playback of either continuous recording of the camera output on analog tape (1), video tape (16), or successive frames of digital tape (3) without data processing. These techniques are applicable when either a limited number of sites are required, when the combination of counting rate and accumulation time will yield the desired statistical accuracy, or when, as in most applications of functional scanning, a large number of elements are required so as to visualize an entire organ, or when it is necessary that the desired parameter be derived by some process such as least squares fitting, a computer of appreciable capabilities must be considered as an integral part of the functional scanning system.

ACKNOWLEDGMENTS

Miss Elaine A. Roth and Mr. William P. Drescher have been of great assistance in the above studies, both in patient measurement and in formulation of the method of computer analysis and display.

The computer facilities were provided by the George M. Humphrey Cardio-pulmonary Laboratory of University Hospitals of Cleveland.

REFERENCES

1. W.E. Adam, P. Schenck, H. Kampmann, W.J. Lorenz, W.G. Schneider, W. Ammann, and L. Bilaniuk, "Investigation of Cardiac Dynamics using Scintillation Camera and Computer," Medical Radioisotope Scintigraphy, Vol. II, 77, (International Atomic Energy Agency, Vienna, 1969).
2. M.A. Bender, L. Moussa-Mahmoud, and M. Blau, "Quantitative Radiocardiography with the Digital Autofluoroscope," Medical Radioisotope Scintigraphy, Vol. II, 57, (International Atomic Energy Agency, Vienna, 1969).
3. E.M. Sigman, M.A. Bender, and M. Blau, "Radiohippuran Renography and Radiohippuran Renal Autofluoroscopy," J. Urol., 92, 153 (1964).
4. M.K. Loken, R.E. Linnemann, and G.S. Kush, "Evaluation of Renal Function using a Scintillation Camera and Computer," Radiology, 93, 85 (1969).
5. J.R. Medina, J.P. Lilliehe, M.K. Loken, and R.V. Ebert, "Use of the Scintillation Anger Camera and Xenon Xe^{133} in the Study of Chronic Obstructive Lung Disease," J.A.M.A., 208, 985 (1969).
6. S.R. Inkley and W.J. MacIntyre, "Dynamic Lung Scanning with the use of $^{133}Xenon$," J. Lab. and Clin. Med., 74, 885 (1969).
7. C.T. Dollery, P. Hugh-Jones, and C.M.E. Matthews, "Use of Radioactive Xenon for Studies of Regional Lung Function: A Comparison with Oxygen," Brit. Med. J., 15, 1006 (1962).
8. W.J. MacIntyre and S.R. Inkley, "Functional Lung Scanning with $^{133}Xenon$," J. Nucl. Med., 10, 355 (1969).
9. S. Kaihara, T.K. Natarajan, H.N. Wagner, Jr., and C.D. Maynard, "Construction of a Functional Image from Regional Rate Constants," J. Nucl. Med., 10, 347 (1969).
10. W.J. MacIntyre, S.R. Inkley, E. Roth, W.P. Drescher, and Y. Ishii, "Spatial recording of Disappearance Constants of $^{133}Xenon$ Washout from the Lung," J. Lab. Clin. Med., 76, 701 (1970).
11. W.J. MacIntyre and S.R. Inkley, "Evaluation of Regional Lung Function with $^{133}Xenon$," (in press), Radioaktive Isotope in Klinik und Forschung, Vol. 10, (München-Berlin, Urban und Schwarzenberg, 1973).
12. E. Oberhausen, K.J. Neumann, and W. Schiffler, "Method for Adjusting the Anger Camera," Medical Radioisotope Scintigraphy, Vol. I, 145, (IAEA, Vienna, 1969).
13. W.J. MacIntyre, J.H. Christie, and G.S. Curtis, "Three-dimensional Computer Read-out of Radioisotope Scan Data," Radiology, 90, 22 (1968).

14. W.J. MacIntyre and J.H. Christie, "A comparison of data Averaging of Radioisotope Scan Data by Photographic and Dimensional Computer Techniques," Medical Radioisotope Scintigraphy, Vol. I, 771, (International Atomic Energy Agency, Vienna, 1969).
15. G. Miörner, "¹³³Xe-Radiospirometry: A Clinical Method for Studying Lung Function," Scan. J. Resp. Dis., Suppl. No. 64 (1968).
16. W.L. Ashburn, J.C. Harbert, W.C. Whitehouse, and D.T. Mason, "A Video System for Recording Dynamic Radioisotope Studies with the Anger Scintillation Camera," J. Nucl. Med., 9, 554 (1968).

A Proposed Method for Quantitative Brain Blood Flow Transverse-Section Scanning

J. A. Patton, A. B. Brill, and P. H. King

Abstract

Many approaches to tomographic imaging of static distributions have been reported in the literature. However, tomographic imaging of changing distributions is much more difficult. A study is being conducted to determine the feasibility of transverse-section tomographic imaging of blood flow through the brain. The method utilizes a cylindrical array of 8 sodium iodide detectors. In a typical scanning procedure, the field of view of each detector is swept through the scanning area by a precision stepping motor and timing belt assembly. Counts versus angular position of each detector are recorded separately in memory. Multiple frames would be recorded by multiple passes of the detectors. The data are transferred to a small computer for image construction which requires less than one minute of computer time for each image. Good quality images of static distribution have been recorded in less than one minute of scan time. The gearing ratios are being changed and the drive system is being improved to permit frame rates on the order of 1 to 2 seconds. Studies made with this improved drive mechanism should provide valuable information on the merits of transverse-section imaging of changing distributions.

**Vanderbilt University
Nashville, Tennessee**

**The work was supported by American Cancer Society
Research Grant #ET-21**

A Proposed Method for Quantitative Brain Blood Flow Transverse-Section Scanning

J. A. Patton, A. B. Brill, and P. H. King

Transverse section tomography has been under study in our laboratory for the past three years (1, 2, 3). An eight detector cylindrical array tomographic scanner optimized for brain scanning has been constructed and evaluations are now under way. A typical scanning situation with this system is shown in Figure 1. The eight detectors are aimed at a common scanning point which is directly below the driving point of the system. The eight detectors are mechanically coupled to the driving point and thus are forced to follow the scanning point as it is driven in a rectilinear raster by a computer controlled x-y plotter. Summed counts for the eight detectors versus position of the common scanning point are collected in an array of 64 by 64 data points corresponding to an 8 by 8 inch field of view. The data are displayed in an array of 256 by 256 display points with 64 shades of gray for photographic purposes on a CRT attached to the computer. The display array is constructed from the smaller data array by linear interpolation in both the x and y directions between adjacent data points (4). Ten minutes of scanning time are required to collect images from patients.

This raster method of data collection has one distinct disadvantage which is inherent in all of the other tomographic systems that have been reported. This disadvantage is the inability to collect data at a rate high enough to map changing distributions. Another method of data collection is being studied with the V. U. Tomo Scanner, denoted as the constant angular velocity method, which may make it possible to obtain dynamic function studies. A photograph of the top of the scanner in this mode of operation is shown in Figure 2. In this procedure, data are collected at preset angular increments from each detector as its field of view is swept through the scanning area at a constant angular velocity by a precision stepping motor and timing belt assembly. Data from only a single pass are required for construction of an image. The data are stored until completion of the scan and then the image is constructed by means of computer programs.

A block diagram of the data acquisition system is shown in Figure 3. On the logic control panel, switches are available for setting the speed, sampling interval, number of data points, and direction. Individual data lines from the 8 detectors are brought into the interface and connected to 8 eight-bit binary data registers. After the desired settings have been selected on the control panel, a start button is pressed which sends pulses to the stepping motor drive system, and also starts data collection with the 8 data registers. After the number of pulses corresponding to the desired angular increment have been sent to the drive system and counted in the pulse counting register, this register is reset, data collection is stopped, the 8 data registers are multiplexed into 8 separate sections of core in the 4096 word memory and reset, and data collection begins again.

FIG. 1

FIG. 2

FIG. 3

This procedure is repeated until the proper number of data points have been collected to construct an image of the activity distribution being examined. At the completion of a scan, the data are transferred to a PDP-9 computer for image construction.

The image to be constructed is a 64 x 64 data point array representing an 8 x 8 inch square field of view. In order to simplify calculations and computer manipulations all measurements are made in units of 0.125 inches which is the distance represented by adjacent data points in the image array. Data are collected in angular increments of 1 degree from -34 degree to +34 degrees which will just include the 8 x 8 inch array as shown in Figure 4. Thus, for each detector, there will be 69 data points collected to be used to construct a 4096 data cell array. The image array for each detector is constructed by using a 64 x 64 array of angular factors which is stored on the disc and generated in the following manner. The initial angle corresponding to the data cell (1,1) in the array is -34 degrees with the origin of the coordinate system taken as the center of rotation of the detector in question. The distance from the origin to the closest point in the 4096 array (33,1) is $OC=48$ (6 inches). The distance to the most distal point in the array is $OB=122$ (15.25 inches). Thus, referring to the array in polar coordinates, the limits on θ are $(-34, 34)$ and on r are $(48,122)$.

The initial conditions are therefore $\theta = -34$ degrees and $r = 48$. This point is converted to rectangular coordinates, the origin is translated to the position corresponding to data cell (1,1) and a check is made to see if this position point falls within the 64 x 64 array. The x and y coordinates are rounded off to the nearest whole number by adding 0.5 to each coordinate and truncating it in the computer. If this position point falls within the array, the angular value is entered into the corresponding point in the array. If this position does not fall within the array, the radius is incremented by 1 (0.125 inches) and the process is repeated. This procedure continues until the radius reaches its maximum value of 122 (15.25 inches). θ is then incremented by 1 degree corresponding to the next collected data point, r is reset to 48, and the process continues until the entire set of 69 collected data points has been expanded into an array of 64 x 64 angular values which may be used to construct an image for a single detector.

An image generated from a single detector with simulated point source data is shown in Figure 5A. From this image it can be seen that the entire array is not filled by the image generation. These blank data cells are filled in by linear interpolation in the x direction between adjacent non-zero points by a sub-routine titled FILLIN. The results of the interpolation are shown in Figure 5B.

After the array corresponding to each detector is constructed it is stored on the computer disc. The arrays are then summed together in the following manner. Referring again to Figure 5, the first array is rotated 90 degrees and added to array number three. This summed array is rotated 90 degrees and added to array number five. This summed array is again rotated 90 degrees, added to array number seven, and stored on the disc. The same procedure is repeated for arrays two, four, six, and eight. A sum of arrays for four orthogonal detectors is shown in Figure 5C. For the final summation a 45 degree rotation is required of the sum of arrays one, three, five, and seven. This rotation is performed by rotating each point about the center (32.5, 32.5) of a zeroed 64 x 64 data point array, rounding off each coordinate to the nearest whole number, and checking to see if the rotated point falls within the zeroed array. If the point lies

$AB = BE = 64$ (8 IN.)
 $OD = 80$ (10 IN.)
 $OC = 48$ (6 IN.)
 $OB = 122$ (15.25 IN.)
 $\theta = 34^\circ$

$$\begin{aligned}
 X &= R \sin \theta \\
 Y &= R \cos \theta
 \end{aligned}$$

FIG. 4

A. SINGLE DETECTOR
IMAGE

B. (A) WITH FILLIN

C. SUM OF FOUR
ORTHOGONAL DETECTORS

D. (C) ROTATED 45
DEGREES

E. (D) WITH FILLIN

F. FINAL IMAGE

FIG. 5

within the array, it is inserted into the proper position. This 45 degree rotation is actually performed using an array of rotational factors which are stored in the disc. The result of this 45 degree rotation is shown in Figure 5. Again, every cell in the array is not filled. Therefore, the FILLIN subroutine is used again on this array as shown in Figure 5E. This array is then added to the summed array from detectors two, four, six and eight to produce the final image as shown in Figure 5E. This image is the same in appearance as those that have been obtained with the raster method by simply summing the outputs of the 8 detectors. The star effect which is inherent in the transverse-section tomography may be masked as in the summing mode by using a baseline cut-off procedure.

Data were collected in the constant angular velocity (CAV) mode from a single point source, four point sources, and a phantom in scanning situations identical to those that have been used for the raster mode. During image construction, an alignment problem was discovered. This problem was attributed to stretching of the PIC timing belt and flexing of the teeth in the belt. It was also observed that this effect occurred primarily at the beginning of the scan so that during most of the scan the detectors maintained their proper relation between adjacent angular increments. To correct for this problem, a CAV scan was collected from a point source positioned precisely in the center of the scanning area. By examining the difference between the actual center of the set of data from each detector and the measured center using the point source, a shift factor was determined for each detector. This factor was then used to shift each data set collected so that the proper orientations were obtained before image construction was attempted.

Images constructed from data collected in the CAV mode and shifted by using the shift procedure are shown in Figure 6. They were collected in about one minute of scan time as compared to eight minutes for the raster mode data. The activities of the sources in each instance were decreased by a factor of eight for the CAV study with no loss in image quality. This overall reduction in scan time verifies the advantages of the CAV mode which we anticipated by removing the redundancy of the raster mode. By examining these scans and comparing them to those collected in the raster mode, no differences can be detected in the image quality obtained with the two modes of operation. Thus, with the CAV mode, scans of image quality comparable to those obtained in the raster mode are obtainable in about 1/60 of the scan time. Image construction requires 1 minute of computer time on a small computer.

An evaluation of the CAV mode for data collection from patients is now being planned. In order to remove the alignment problem due to the stretching of the drive belt, this drive system is being replaced with precision 3/8 inch timing belts and timing gear assemblies. Idler pulleys will be placed in strategic locations to remove any excess play in the belts. The gearing ratios are also being changed to permit collection times of 1 to 2 seconds for a single image. This will permit us to study the usefulness of transverse-section tomography in studying cerebral blood flow.

A. POINT SOURCE

B. 4 POINT SOURCES

C. PHANTOM

FIG. 6

Since the data are maintained separately from each of the detectors, we are studying the use of computer manipulation and rejection techniques performed with the goal of removing the star effect and enhancing the final image obtained. Another idea that warrants further study with this system is the use of the collimator response characteristics to distribute counts along each ray constructed from a particular detector. An event recorded by a detector most probably originated at the focal point of the collimator which is the point of maximum sensitivity and resolution. The probability of occurrence falls off on either side of the focal point. Thus it should be possible to use the response characteristics of the collimators to define a probability function to be used to distribute throughout the field of views, counts obtained from a particular detector at a particular angular position. It is not known at this time whether significant improvement in image quality will be obtained using this procedure, but the idea has enough promise to warrant further study.

Preliminary studies with this system should determine the merits of transverse-section mapping of cerebral blood flow in general, and in particular, the merits of cylindrical arrays of detectors for this type of imaging. If this procedure provides useful data, we may add additional rings of detectors to obtain three-dimensional views of changing isotope distributions. It may be feasible to use a smaller number of detectors in each level to permit closer packing of adjacent levels. One could then use three-dimensional spatial averaging to reduce the effects of statistical fluctuations. Studies are planned to evaluate these concepts.

Acknowledgment:

This work was supported by American Cancer Society Research Grant # ET-21.

References

1. J.A. Patton, A.B. Brill, J.J. Erickson, W.E. Cook, and R.E. Johnston. "A New Approach to the Mapping of Three Dimensional Radionuclide Distributions," Journal of Nuclear Medicine, 10, 6, (1969), p. 363.
2. J.A. Patton, A.B. Brill, J.J. Erickson, and P.H. King, "Cylindrical Array Tomographic Scanner with Focused Collimators and Its Possible Adaptation for Fluorescence Scanning of Brain Tumors," Southern Medical Journal 62, 11, (1969), p. 1434.
3. A.B. Brill, J.A. Patton, J.J. Erickson, and P.H. King. Multicrystal Tomographic Scanner for Mapping This Cross-Section of Radioactivity in an Organ of the Human Body. Patent No. 3,591,806, Assigned to the U.S.A. by the U.S.A.E.C., July 6, 1971.
4. J.A. Patton, J.J. Erickson, and A.B. Brill. "Optimization of Array Size for Display Purposes," Southern Medical Journal 63, 11, (1970) p. 1372.

Figures

1. Clinical Scanning Situation with the V.U. Tomo Scanner.
2. Photo of top of scanning system with constant angular velocity drive mechanism.
3. Block diagram of the data acquisition system for data collection in the constant angular velocity mode of operation.
4. Reference data for construction of image from Tomo Scanner by computer.
5. Image construction with simulated point source scan data.
6. Scans collected in constant angular velocity mode and displayed with 60% background subtraction or baseline cutoff.

THE HEIDELBERG SCINTIGRAPHIC IMAGE PROCESSING SYSTEM

by

P. Pistor*, P. Georgi**, and G. Walch*

ABSTRACT

A software package (operational on an IBM 360/50) for processing Anger camera data and scanner data are described. The concepts of the processing system and the computational methods used for filtering are discussed. Some examples of processed clinical scintigrams are presented.

(TWA)

* IBM Wissenschaftliches Zentrum Heidelberg

** Institut für Nuklearmedizin im Deutschen
Krebsforschungszentrum

1. INTRODUCTION

Scintigraphic data should not only be recorded as pictures (e.g. scintiphotos, dot scans), but also in such a way that the measurements can be redisplayed with different parameters. The records may be analog or digital for that purpose. However, digital records should be preferred as they enable one to encode the image information so that statistical fluctuations (image noise) are suppressed, while contrast is increased or at least preserved.

This paper describes the concepts of a software package which has been operational for more than one year on the /360-50 digital computer of the IBM Scientific Center Heidelberg. It processes scintigraphic data which are recorded off-line at the German Cancer Research Center and at the radiological clinic of the Heidelberg University.

2. DESCRIPTION OF THE PROCESSING SYSTEM

A block diagram of the processing system is given in fig. 1. It can be used to process Anger camera data and scanner data, which are both available in digital form. The problem of their acquisition will not be discussed here.

As outlined below the data flow is different for different types of data.

2.1 Flow of Anger Camera Data

The data consist of coordinates of γ -quanta, sequentially stored on magnetic tape according to the time of their detection (sequential mode). The raw data are filed and then used to build up a matrix of 256×256 cells, representing a two dimensional discrete count-rate distribution. Further processing is different for image data and camera system data.

2.1.1 Camera System Data

Two kinds of camera system data are recorded, the responses to flat phantoms and to point sources. Flat phantom data are used to correct for the non-uniform plane sensitivity of the Anger camera¹; point source data are needed to compute filters.

2.1.1.1 Flat Phantom Response

The flat phantom response is first smoothed columnwise and rowwise. The smoothing operator has a Gaussian shape and a full width at half maximum (FWHM) of about 5 cell widths. Usually the smoothed data are compressed to a 128×128 matrix by condensing four neighbouring cells into one. It can be shown (e.g. Hunt², Metz³) that this procedure results in an additional smoothing.

Fig. 1: Block diagram of the processing system
 (...: off line loop, ---: not yet implemented)

The smoothed and compressed data are filed in digital (tape, disk) and analog (grey level print) form.

2.1.1.2 Point Source Response

Usually the response to several point sources located at different positions is measured. The data are eventually compressed, corrected for the flat phantom measurements of the corresponding isotope, and then stacked. The last procedure is performed to obtain an average point source response* applicable for the major part of the field of view of the camera. To reduce the influence of truncation the stacked point source response is multiplied by a weighting function (tapering) which is constant in the center and decays gently at the edges of the response function. The lateral extension necessary for a correct record is found by visual inspection of printer displays.

The final point source response is stored on disk or tape for further use.

2.1.1 Image Data

Image data (compressed or not compressed) are filed after correction for the flat phantom. The uncorrected data are eventually stored too for reprocessing and for comparison purposes. Only corrected data are filtered.

2.2. Flow of Scanner Data

The flow diagram for scanner data is less complicated than for Anger camera data. Both image data and point source response data are obtained in matrix form and need not be rearranged from sequential records. As the plane sensitivity of a scanner is constant, corrections for flat phantom measurements are not necessary. In addition the point source response is independent of the source location (provided the image cells are small enough); thus the response to a single source is sufficient to describe the system.

2.3 The Filter Module

Data preprocessed for filtering are divided into two classes only: image data and point source response data (see fig. 1). We will describe now in some detail how these data are used to set up a filter and to produce processed image data.

2.3.1 Concepts of Filter Design and Filter Computation

To understand which data are needed to construct the filter used, let us consider a 3-dimensional activity density distribution $\rho(x,y,z)$. It consists of an infinite layer of constant thickness H and constant activity density ρ_1 , which contains regularly spaced volumes of simple geometrical shape (e.g. spheres or cylinders) and

of constant activity density ρ_2 ($\rho_2 < \rho_1$: cold spots, $\rho_2 > \rho_1$: hot spots). Activity distributions better adapted to clinical reality may be embedded as indicated in fig. 2.

Fig. 2: artificial 3-d activity density distribution (cross section)

$\rho(x,y,z)$ is condensed with respect to z to obtain a 2-dimensional activity density distribution $o(x,y)$, the so called object distribution:

$$(1) \quad o(x,y) = \int_0^H \rho(x,y,z) dz.$$

It is depicted in figure 3.

Fig. 3: artificial object distribution (cross section)

Obviously the quantities Δo and Δo of fig. 3 are given by

$$(2) \quad \Delta o = \rho_1 H$$

and

$$(3) \quad \Delta \tilde{o} = (\rho_2 - \rho_1) h,$$

respectively. $o(x,y)$ is a constant Δo , modulated by

$$(4) \quad \tilde{o}(x,y) = o(x,y) - \Delta o.$$

The object modulation is composed of elementary object details $o_{det}(x,y)$ as shown in fig. 4.

Fig. 4: object modulation and elementary object detail (cross section)

Let $b(x,y)$ * denote a scintigram and s a point spread function, that is a point source response renormalized such that

$$(5) \quad \iint_{-\infty}^{+\infty} s(x,y) dx dy = 1.$$

It is assumed that the expectation of b is given by the convolution of o and s :

$$(6) \quad E(b(x,y)) = \iint_{-\infty}^{+\infty} o(x',y') s(x-x',y-y') dx' dy'.$$

According to equation (6) $\tilde{b}$, the scintigram without the constant component obeys

$$(7) \quad E(\tilde{b}(x,y)) = \iint_{-\infty}^{+\infty} \tilde{o}(x',y') s(x-x',y-y') dx' dy'.$$

---

* For sake of simplicity all functions are considered to be functions of continuous variables (x,y) rather than discrete variables $(i\delta x, j\delta y)$. The number b_{ij} of events counted in an image cell (i,j) of the size $\delta x \cdot \delta y$ is given by

$$\int_{x_i - \delta x/2}^{x_i + \delta x/2} \int_{y_i - \delta y/2}^{y_i + \delta y/2} b(x,y) dx dy$$

The b_{ij} are Poissonian random variables, and are statistically independent of b_{kl} if (i,j) and (k,l) denote different cells.

From $\tilde{b}(x,y)$, $\tilde{o}(x,y)$ is estimated by the convolution

$$(8) \quad \tilde{c}(x,y) = \iint_{-\infty}^{+\infty} \tilde{b}(x',y') f(x-x',y-y') dx' dy'$$

in such a way that the quantity

$$(9) \quad I = \lim_{a \rightarrow \infty} a^{-2} \int_{-a/2}^{a/2} \int_{-a/2}^{a/2} (\tilde{c}(x,y) - \tilde{o}(x,y))^2 dx dy$$

is minimized (Wiener criterion). It can be shown⁴ that the filter which meets this condition (Wiener filter) has the Fourier transform

$$(10) \quad F(u,v) = \frac{1}{S(u,v)} \frac{|O_{\text{det}}(u,v)|^2 |S(u,v)|^2}{|S(u,v)|^2 |O_{\text{det}}(u,v)|^2 + N},$$

where $O_{\text{det}}(u,v)$ and $S(u,v)$ are the Fourier transforms of $o_{\text{det}}(x,y)$ and $s(x,y)$, respectively, and N is given by

$$(11) \quad N = E\left\{\int_0^{\Delta x} \int_0^{\Delta y} b(x,y) dx dy\right\}.$$

Δx and Δy denote the separation of the object details in $o(x,y)$ (see fig. 3).

The quantities entering equation (10) are known either by measurement or manufacturers' specifications ($s(x,y)$), or by the model described above ($\rho_1, \rho_2, h, H, \Delta o, \Delta \tilde{o}, o_{\text{det}}(x,y)/\Delta \tilde{o}, \Delta x, \Delta y, N$). It is an important question how to specify the parameters and functions of the model in order to adjust it to given scintigrams. For instance, if the (cold/hot) spots are assumed to be spheres, $o_{\text{det}}(x,y)/\Delta \tilde{o}$ is uniquely defined by the diameter h . The fraction

$$(12) \quad \frac{\Delta \tilde{o}}{\Delta o} = \frac{\rho_2 - \rho_1}{\rho_1} \frac{h}{H}$$

requires assumptions about the storage anomalies and the surrounding tissue, namely the activity density contrast $(\rho_2 - \rho_1)/\rho_1$ and the geometrical proportion h/H . If these fractions are small, the parameters $\Delta o, \Delta x, \Delta y$, and N are approximately interrelated by

$$(13) \quad \Delta o \cdot \Delta x \cdot \Delta y = N.$$

N is estimated from a given scintigram by the total number of γ -quanta detected in a region of interest covering an $(\Delta x \cdot \Delta y)$ -area. Δx and Δy are arbitrary parameters. Good results have been obtained if they are chosen such that

$$(14) \quad \int_{-\Delta x/2}^{+\Delta x/2} \int_{-\Delta y/2}^{+\Delta y/2} b^2_{\text{det}}(x,y) dx dy = 0.999 \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} b^2_{\text{det}}(x,y) dx dy,$$

where b_{det} , the image of the object detail, is given by

$$(15) \quad b_{det}(x,y) = \iint_{-\infty}^{+\infty} o_{det}(x',y')s(x-x',y-y')dx'dy'.$$

Equation (10) is evaluated using Fast Fourier Transform (FFT) techniques^{5 6}. Computing times are strongly affected by the filter size and by the separability of $s(x,y)$ and $o_{det}(x,y)$, that is

$$(16) \quad o_{det}(x,y) = o_1(x) \cdot o_2(y),$$

$$(16') \quad s(x,y) = s_1(x) \cdot s_2(y).$$

If equations (16) hold, then 128 x 128 samples of $F(u,v)$ are computed in about one second on an IBM /360-50. The equations (16) are valid, e. g., if the functions have a Gaussian shape. For non-separable functions $o_{det}(x,y)$ and $s(x,y)$ the computing effort is increased by one to two orders of magnitude, especially if access to external devices is necessary because of limited core storage capacity.

A typical amplitude-spectrum $|F(u,0)|$ is displayed in figure 5.

Fig. 5: Typical filter response in frequency domain (filter renormalized to $F(0,0)=1$).

Filters with similar frequency response have been obtained, e. g. by Nagai et al⁷, Metz³, Tanaka and Iinuma⁸, Brown et al⁹. All these filters tend to approximate $1/S(u,v)$ for low frequencies. According to their different design philosophies they mainly differ in the frequency of maximum amplitude and in the transition to the stop band.

By the imaging process, o_{det} is transformed to a smeared version b_{det} , and the contrast $\Delta\tilde{O}/\Delta o$ is reduced to $\max(b_{det})/\Delta o$. In addition all features are obscured by statistical noise having an approximately

constant variance $\sigma^2 = \Delta o$ (fig 6).

Fig. 6: Visualization of image degradation and image enhancement.

Now the filter is expected to produce an output

$$(17) \quad c_{det}(x,y) = \iint_{-\infty}^{+\infty} b_{det}(x',y')f(x-x',y-y')dx'dy'$$

(processed object detail) which first appears to be compressed in comparison to b_{det} , secondly as an improved contrast

$$(18) \quad \frac{\max(c_{det}) - \min(c_{det})}{\Delta o'} = \frac{\Delta c_{det}}{\Delta o'} > \frac{\max(b_{det})}{\Delta o},$$

$$(19) \quad \Delta o' = \Delta o \cdot F(o,o),$$

and thirdly has an improved signal to noise amplitude ratio

$$(20) \quad \frac{\Delta c_{det}}{\sigma'} > \frac{\max(b_{det})}{\sigma}$$

where

$$(21) \quad \sigma'^2 = \Delta o \iint_{-\infty}^{+\infty} f^2(x,y)dx dy = \Delta o \iint_{-\infty}^{+\infty} |F(u,v)|^2 du dv$$

is the marginal variance of the filtered noise.

The quantities c_{det} , Δc_{det} , and σ'^2 can be used to characterize the performance of a filter, and therefore they are computed in addition to $F(u,v)$. Together with the design parameters they are denoted as control information in fig. 1.

2.3.2 Filter Application

Making use of the so called convolution theorem the image data are not filtered according to equation (8), but by application of the frequency filter (10). The procedure is depicted in fig. 7.

Fig. 7: Block diagram of frequency filtering.

The filter output is stored together with several parameters characterizing the filter performance.

4 Interactive Display Module

To display unprocessed or processed digital image data grey level prints are used. As the number of symbols is limited, it is important to display only interesting intensity intervals. They are chosen by trial and error using an IBM 2250 display unit. This device has been programmed to give access to image files, and to display frames of 2×52 cells with ten intensity levels. The image field is used as a window to visualize detail or total views scaled down by factors $1/2$, $1/3$, or $1/4$.

Each time an image is accessed a standard intensity interval is displayed first. The cut-off levels may then be changed by alphanumeric keyboard, by programmed function keys or by the extreme intensities of a region of interest marked by light pen.

As the intensity resolution is limited, an additional investigation of high resolution cross-sections should support conclusions drawn from 2-dimensional intensity displays. The system displays horizontal and vertical cross-sections which are selected by light pen.

Printer displays of both curves and images are initiated at the display unit. The images, detail views or total views, are printed with the cut off levels of the screen display.

5 Implementation of Other Interactive Facilities

The interactive display unit could be used to change filter parameters to control their influence on both the filter output and the filter performance information. A further possibility is the selection of regions of interest for dynamic function studies based on sequential data. These facilities are indicated in Fig. 1 but have not yet been implemented.

3. EXAMPLES OF PROCESSED CLINICAL SCINTIGRAMS

Up to now about 600 scintigrams have been processed by the Heidelberg System. Some examples of Anger camera pictures are given in this chapter.

Figure 8 shows a head scintigram with multiple metastases. The female patient has been treated for mamma carcinoma two years before this measurement.

The next figure represents a breast scintigram of a female patient with osteogeneous sarcoma in the left knee. A large metatasis is visible in both the unprocessed and processed image, a small one just below it is visible only in the processed scintigram. The intensity range displayed in figure 8b emphasizes the weak anomaly.

The figures 10 and 11 show a frontal and a lateral view of a liver with a metastasis in the upper part of the organ. The first view is also given as it appears on the 2250 display unit (fig. 12). Orthogonal cross-sections through the cold spot of the frontal view are displayed in fig. 13. The curves referr to unprocessed (fig. 10a) and processed (fig. 10b) data and demonstrate the influence of the filter.

FIG. 8a

FIG. 8b

FIG. 9a

FIG. 9b

FIG. 10a

FIG. 10b

FIG. 11a

FIG. 11b

FIG. 13b

FIG. 13a

FIG. 12

ACKNOWLEDGEMENTS

This paper reports about results of a study jointly carried out by the Institute of Nuclear Medicine of the Deutsches Krebsforschungszentrum and by the Scientific Center of IBM Germany. The authors wish to thank Professor K. E. Scheer and Dr. W. J. Lorenz as well as Dr. W. Kattwinkel and Dr. H. G. Meder for their support and encouragement.

Mrs. M. Gamp and Mrs. R. Sindlinger typed the manuscript, and Mr. J. Schmitt prepared the drawings. Mr. M. Bergen read the paper, and Mr. U. Willm was responsible for data preprocessing. The authors gratefully acknowledge their cooperation.

REFERENCES

1. W. Ammann, W. E. Adam et al, Der Einsatz eines Computers in der Szintigraphie, Rad. Biol. Ther., 9(6), 703 (1968).
2. W. A. Hunt, H. G. Meder et al, Optimum Sample Size in Digital Radioscintigraphy, IBM Wiss. Zentrum Heidelberg - 70.09.004 (1970).
3. C. E. Metz, A Mathematical Investigation of Radioisotope Scan Image Processing, (University of Pennsylvania, Ph.D. Thesis, 1969).
4. P. Pistor, G. Walch et al, Digital Image Processing in Nuclear Medicine - Methods and Results -, IBM Wiss. Zentrum Heidelberg - 71.05.002 (1971).
5. W. T. Cochran et al, "What is the Fast Fourier Transform?", Proc. IEEE, 55, 1664-1674 (Oct. 1967).
6. N. M. Brenner, Cooley - Tukey Fast Fourier Transform - Four2 in Assembler Language, IBM Contributed Program Library 360 - 13.4.008 (1969).
7. T. Nagai, T. A. Iinuma and S. Koda, "Computer Focusing for Area Scans", J. Nucl. Med., 9, 507-516 (1968).
8. E. Tanaka and T. A. Iinuma, "Approaches to Optimal Data Processing in Radioisotope Imaging", Phys. Med. Biol., 15, 683-694 (1970).
9. D. W. Brown, D. L. Kirch et al, "Computer Processing of Scans Using Fourier and Other Transforms", J. Nucl. Med., 12, 287-291 (1971).

FIGURE CAPTIONS

- Fig. 8 Head scintigram, taken 2h p.i.
(a) unprocessed data, (b) processed data.
Isotope: ^{18}F , 6mC.
Maximum number of counts/cell: 58(1400/cm²) (fig.a).
Linearly divided intensity range: 20-85 % (fig.a),
15-80 % (fig.b).
- Fig. 9 Breast scintigram, taken 2h p.i.
(a) unprocessed data, (b) processed data.
Isotope: ^{18}F , 7mC.
Maximum number of counts/cell 73(1800/cm²) (fig.a).
Linearly divided intensity range: 15-80 % (fig.a),
10-40 % (fig.b).
- Fig. 10 Liver scintigram, frontal view, taken 20 min.p.i.
(a) unprocessed data, (b) processed data.
Isotope: ^{198}Au , 150uC.
Maximum number of counts/cell 78(1600/cm²) (fig.a).
Linearly divided intensity range: 25-85 %.
- Fig. 11 Same as fig. 10, right lateral view.
Maximum number of counts/cell: 82(1600/cm²)
Linearly divided intensity range: 20-80 % (fig.a),
40-90 % (fig.b).
- Fig. 12 Same as fig. 10, screen display.
- Fig. 13 Same as fig. 10, intensity cross-sections
(a) unprocessed scintigram, (b) processed scintigram.

Preliminary Results on the Use of the Hadamard Transform

C. Berche, A.E. Todd-Pokropek* and R. Di Paola

Institut Gustave-Roussy
16 bis. Avenue P. Vaillant Couturier
94-Villejuif
France

* also of

Institute of Nuclear Medicine
The Middlesex Hospital Medical School
Mortimer Street
London W.1.
England

ABSTRACT

A short exposition of the theory of the Hadamard transform is given. An alternative method of filtering for use in the Hadamard domain, which also works in the Fourier domain is shown with examples. It is suggested that as a result of the very considerable increase in speed resulting from working in the Hadamard domain, this technique may be of some value.

Preliminary results on the use of the Hadamard transform

It is necessary in introducing the results that have been obtained using the Hadamard transform to include a little of the theory that lies behind, while noting that although the theory of the Fourier transform is well-established, the theory of the Hadamard transform is complex and still in course of development [1,2,3,4].

The Hadamard transform is essentially a decomposition into trains of square waves, in contrast to the Fourier transform, which is a decomposition into sinusoids. The square waves take only values of ± 1 , and as a result the transform is particularly simple since it involves only additions and subtractions, as opposed to the multiplications required in the Fourier transform. It is also convenient and easy to work with integer, rather than a floating point, arithmetic. An analogue to the fast Fourier transform, called the fast Hadamard transform [5,6,7], also exists, which is particularly simple to program when working with matrices which are an exact power of 2 in size.

The bi-dimensional Hadamard transform may be defined as

$$TH \{A_n\} = H_n \cdot A_n \cdot H_n$$

where H_n is a Hadamard matrix of order n and A_n is the image matrix of the same size. This is similar to the definition of the Fourier transform. H_1 , the smallest Hadamard matrix, is:

$$\begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix}$$

H_2 , the matrix of size 4×4 , may be defined as the Kronecker product ($\otimes$) of H_1 with itself, which gives

$$\begin{aligned} \begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} \otimes \begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} &= \begin{bmatrix} 1 \times \begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} & 1 \times \begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} \\ 1 \times \begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} & -1 \times \begin{bmatrix} 1 & 1 \\ 1 & -1 \end{bmatrix} \end{bmatrix} \\ &= \begin{bmatrix} 1 & 1 & 1 & 1 \\ 1 & -1 & 1 & -1 \\ 1 & 1 & -1 & -1 \\ 1 & -1 & -1 & 1 \end{bmatrix} \end{aligned}$$

H_2 , the matrix of size 8×8 , is similarly defined as $H_2 \otimes H_1 \otimes H_1$, and H_n as the Kronecker product of H_1 with itself n times.

The fast Hadamard transform results from the following property: It is possible to define a matrix H'_n such that:

$$\left(H'_n \right)^n = H_n$$

where H_n is defined as above.

Thus, the Hadamard transform may be redefined as:

$$TH\{A_n\} = \left(H'_n \right)^n \cdot A_n \cdot \left(H'_n \right)^n$$

This is of course only valuable if H'_n is sparse.

For n equals 3, the form of H'_n is:

$$H'_3 = \begin{bmatrix} 1 & 1 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 1 & 1 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1 & 1 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 1 & 1 \\ 1 & -1 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 1 & -1 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 1 & -1 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 1 & -1 \end{bmatrix}$$

And in general:

$$\begin{bmatrix} 1 & 1 & & & & & & \\ & & 1 & 1 & & & & \\ & & & & 1 & 1 & & \\ & & & & & & \dots & \\ & & & & & & & \dots \\ & & & & & & & & 1 & 1 \\ 1 & -1 & & & & & & & & \\ & & 1 & -1 & & & & & & \\ & & & & 1 & -1 & & & & \\ & & & & & & \dots & & & \\ & & & & & & & \dots & & \\ & & & & & & & & 1 & -1 \end{bmatrix}$$

with zeros elsewhere.

The matrix multiplication of H_n involves only 2^{n-1} additions and 2^{n-1} subtractions, and therefore the complete bi-dimensional transform requires $2N \log_2 N$ operations, where $N(=2^n)$ is the size of the matrix.

In addition, a concept similar to that of frequency for the Fourier transform may be developed for the Hadamard transform [8].

"Sequency" is defined as (half) the number of times that a given square wave train passes through the value zero (i.e. See Fig.1).

Consider the r th row or column of the matrix resulting from the Hadamard transform. Unfortunately the value is not directly related to the sequency as defined above.

However, if r is decomposed into the binary number:

$$r_n \cdot r_{n-1} \cdot r_{n-2} \dots r_0$$

and r' , the sequency, is similarly developed into the binary number:

$$r'_n \cdot r'_{n-1} \cdot r'_{n-2} \dots r'_0, \text{ then}$$

then r is related to r' as follows:

$$\begin{aligned} r'_0 &= r_n \\ r'_1 &= r'_0 \oplus r_{n-1} \\ r'_q &= r'_{q-1} \oplus r_{n-q} \end{aligned}$$

where $\oplus$ is the operation "EXCLUSIVE OR", or binary addition module 2.

Using this rule, which is a modification of that of Andrews [9], the matrix resulting from the Hadamard transform may be reordered into increasing values of sequency.

Fig.2 shows the raw data of an image which is to be processed. Fig.3 shows the Hadamard transform of this image where the x axis is in terms of sequency. It will be noted that the Hadamard transform is much more irregular than the Fourier transform. The two domains, frequency and sequency, are not exactly equivalent and it must not be presumed that methods which work in the former are applicable in the latter domain. (Ed. note: Fig. 3 is missing)

Hadamard Filters

The class of filter that has been used extensively in the frequency domain may be called a zonal filter, such that values are 'modified' with respect to their frequency. Nine-point smoothing, Wiener filters, etc. etc., are all of this class. If a similar low pass filter is used in the sequency domain, it has been found (See Fig.4) that the results are usually very poor

FIG. 1

FIG. 2

FIG. 4

since the image is considerably distorted. This technique compares very unfavourably with the classic nine-point smoothing, and is therefore of little interest. This is not in accord with the previously published findings of Brown and Kirsch [10]. However, it may be noted that they were working with a particularly simple phantom and that the images they published were of poor quality. In considering the fluctuating nature of the Hadamard transform this result is not particularly surprising.

A type of filter which has been found to be of greater interest is "threshold" filtering, where values are modified not according to their frequency (or sequency) but according to their amplitude. In this case it is not even necessary to re-order the matrix in terms of sequency, and the transform is correspondingly faster. This type of filter, then, rejects all values greater than some upper limit or less than some lower limit. It has been found, as in the frequency domain, that the structure of the image is determined by the (relatively few) values lying between 2% and 100% of the peak value. Similarly, it has been demonstrated that the noise in the image appears to be uniformly distributed (as in the frequency domain) and that a suitable value for the lower cut-off threshold can be chosen to improve signal to noise ratio. It has also been demonstrated that information related to holes and lumps seems to be contained in the intermediate amplitudes. As an illustration, Fig.5 shows the Hadamard filtered result of the image shown in Fig.2, where the information related to the structure of the image has been almost completely removed, but the hole is still clearly visible. An equivalent result with a similar filter in the frequency domain is shown in Fig.6. It will be noted that the computation time involved was ten times greater in the latter case. It should also be noted that this simplistic two-valued threshold filter could be replaced by a more continuous general filter relating all input values and output values.

Conclusions

The Hadamard transform is interesting because of the computational simplicity and speed. It is particularly suitable for mini-computers. However, conventional (zonal) filtering has not been found to be satisfactory. It seems likely, however, that an amplitude filtering technique can be used, for example, to suppress structural information almost completely. This seems to be at least as efficient as in the frequency domain. Additional zonal filtering could be employed, either in the sequency domain, or in the frequency domain. This might be particularly interesting for high count rate scans, such as liver, brain, etc., where the results of conventional data processing techniques have not been very satisfactory. In addition, the technique may provide a shortcut for use in mimicking filters which in the frequency domain require too much time to be implemented on a small computer system.

FIG. 5

FIG. 6

BIBLIOGRAPHY

1. J. Hadamard, "Resolution d'une question relative aux determinants", Bull. Sci. Math., Ser.2, 17, Part I, 240-246 (1893).
2. H.C. Andrews, Computer Techniques in Image Processing, (Academic Press, New York-London, 1970).
3. M. Koenig and J.P. Zolesio, "Problemes numeriques relatifs aux fonctions de Walsh", C.R. Acad. Sc. Paris, Ser.A, 271, 1074-1077, (23 November 1970).
4. J.L. Walsh, "A closed set of normal orthogonal functions", Ann. J. Math., 45, 5-24, (1923).
5. H.C. Andrews and K. Caspari, "Orthogonal transformations", in Computer Techniques in Image Processing, pp. 105-134, (Academic Press, New York-London, 1969).
6. N. Ahmed and K.R. Rao, "Discrete Fourier and Hadamard transforms", Electronics Letters, 6, (7), 221-224, (April 1970).
7. J.E. Whelchel and D.F. Guinn, The Fast Fourier Hadamard Transform and its use in signal representation and classification, pp. 561-573, Eascon 1968 Convention Record, (1968).
8. H.F. Harmuth, "A generalised concept of frequency and some applications", IEEE Trans. Information Theory, IT-14, (3), 375-382, (May 1968).
9. H.C. Andrews, and W.K. Pratt, "Transform Image Coding" Proceedings of a Symposium Computer Processing in Communications, New York April 1969, J. Fox Editor, Microwave Research Institute Symposium Series Vol XIX no 63-84 (Polytechnic Press of the Polytechnic Institute of Brooklyn, Brooklyn N.Y. 1970)
10. D.W. Brown and D.R. Kirch, "Computer Processing of radioisotope scans using the Fourier and other transformations", J. Nucl. Med., 12, 287-291, (November 1970).

Captions for illustrations for 'Preliminary results on the Hadamard transform'

- Fig. 1. A typical square wave train (Walsh function) of sequence three for the interval $\pm A$.
- Fig. 2. A slice through the two dimensional transform of the image shown in Fig. 3.
- Fig. 3. The raw data from the IAEA phantom with two holes, arrowed.
- Fig. 4. The resulting image after use of a low pass zonal Hadamard filter.
- Fig. 5. The resulting image from a sharp cut-off amplitude Hadamard filter, designed to remove the normal structure while retaining the holes. Increasing density indicates increasing likelihood of holes.
- Fig. 6. The resulting image from a similar filter to that used in Fig. 4, but used in the Fourier domain.

SERIAL SCAN DATA ANALYSIS BY "MINI-COMPUTER"

Robert O. Smith, Margaret S. Boyd,
Patrick H. Lehan and Harper K. Hellems

ABSTRACT

Least squares regression analysis can be applied to serial scans when rate of uptake by an organ is linear. Such linear rates of uptake can be obtained from myocardial scanning when selected isotopes and a continuously decreasing infusion are used. Mini-computers can perform the calculations. A description is provided for each program that has been written for the Hewlett-Packard Model 2115A digital computer. These programs can check the correctness of the spatial data and do 9-point averaging. Other programs calculate the slope of the uptake curves, calculate the Y intercept values and then display the resulting data on cathode-ray tube X-Y displays.

Department of Medicine
University of Mississippi Medical Center
2500 North State Street
Jackson, Mississippi 39216

SERIAL SCAN DATA ANALYSIS BY "MINI-COMPUTER"

Robert O. Smith, Margaret S. Boyd,
Patrick H. Lehan and Harper K. Hellems

Department of Medicine, University of Mississippi
Medical Center, Jackson, Mississippi 39216

Using isotope scanning methods, the heart has proven difficult to image for several reasons. For isotopes that are attracted to the myocardium, if the problem of too high or too low gamma energy can be overcome, then obtaining sufficient counts for the desired scanning periods remains a problem. Ventricular cavities and myocardial motion add to the complexities. The addition of a digital computer and serial scanning techniques has aided in overcoming the problems. A pending publication will describe the overall method and results obtained from mongrel dogs.¹ Computer programs and the recording system have not been described prior to this time.

When isotopes of potassium or rubidium are used to maintain a plateau level in arterial blood,²⁻³ the rate of uptake by the myocardium is a rather linear function for more than 30 minutes. If this plateau level is attained before starting to scan and maintained throughout scanning, then the resulting data can be subjected to linear least squares regression analysis. The slope of the uptake curve can be found by solving for "b".

$$b = \frac{N \sum XY - (\sum X \cdot \sum Y)}{N \sum X^2 - (\sum X)^2}$$

The value of Y when X = 0 (Y intercept) can be found by solving for "I". This can be used as an index to isotope appearance time.

$$I = \frac{\sum Y - b \sum X}{N}$$

In both equations,

X = Scan number (i.e. 1, 2, 3, 4 ....).

Y = Counts detected at a particular location for a particular scan.

N = Total number of scans.

The rectilinear scanning system has been previously described.⁴ The recording system uses dual counters with buffer storage for a permanent record on perforated paper tape. The schematic diagram of the counters, storage and timing circuits are shown in Figure 1. The component modules were government surplus 100 kHz computer logic manufactured by Stromberg-Carlson. The upper and lower detector each has its own 8-bit binary counter. Spatial

interval pulses are obtained from the scanner for each 0.1 inch of detector travel. A switch provides the option of dumping counts each 0.1 or 0.2 inch. Ten microseconds after arrival of the selected spatial interval marker, the collected counts are dumped into storage and counting resumes. When an "end of scan line" signal is received from the scanner, all binary bits are set to one in both counters and two octal 377 characters are punched on tape. When the scanner starts down the next line, the counters are still set and two additional 377's are punched. These are handy visual references on the punched tape and are indicators that the counters are likely functioning. They are also "sign posts" for the computer to use in checking for the correct number of scan lines and data dumps/line. The leader and trailer are blank feed frames. Though the design was specifically for a Soroban Model LP-2 CON perforator, slight changes in the timing circuitry should make it adaptable to other models.

For least squares regression analysis of the acquired data, several programs have been written for the Hewlett-Packard 2100, 2114-5-6 series of digital computers with 8K words of memory. In nearly every instance, the first program used is named "AVG3". The only dimension limit to this program is that the array must not have over 2500 elements. The program requests the number of dumps/scan line and number of scan lines. When an answer is given, one character is read in and checked. If it is a feed frame (Octal 0), a raw data tape is indicated and the operator is queried as to whether it is desired to take off the data from the upper or the lower detector. While the computer is reading in the data, a check is made of the correct number of data dumps/scan line and number of lines. Nine-point averaging is next accomplished in which the center element has a weight of five relative to the others. This is in order that its influence won't be too dilute. A single change will allow a weight of from one to nine. Upon completing the computations, a new tape is punched for one detector with the leader and trailer being a series of characters that are 200 octal. Program "RDATA" will produce a new raw data tape for either the upper or lower detector alone. Its leader and trailer characters are also octal 200's.

When the slope of the uptake curve for each element is desired, the "BANDI" program is used. Maximum array size is 2000. The data input for this program is four or more scans that have been processed by "AVG3" or "RDATA". When the computations are complete, a tape is punched with the "b" values for the array and the computer asks if the intercepts are wanted. If they are, additional processing is accomplished and a second tape is punched with intercept values for the scanned area. The slope values are usually sufficiently smooth but the intercept values are subjected to the "AVG3" program for additional smoothing.

Another available program is "CHECK" which checks raw data tapes for extra or missing data. The program "PRTSN" will list on a teletype all the numbers in an array. "SCLPH" will scale from 0-63 all the elements in a matrix and punch the scaled data. The scale range can be easily modified. For data acquisition from the Nuclear Chicago Pho-Gamma camera, "FOTO8" can be used. Upon receipt of an event, "dead-time" before another can be processed is 16.5 microseconds.

Though the teletype can list all the elements in an array, the "RANDD" program can display a 32 x 32 array on a cathode-ray tube (CRT) X-Y display. Use of intensity modulation for this purpose is described elsewhere in these proceedings.⁵ Difficulty of interpretation was encountered when the slope of

the uptake curves (b's) were on one Polaroid print and the indices to isotope appearance time (I's) were on another. Addition of color has surmounted this difficulty. A double exposure onto the same print is used. The "b's" are photographed through a red filter and the "I's" through a green filter. Wherever both colors impinge, shades of yellow result. Examples of the color readout will be found elsewhere.¹

To obtain more information than could be obtained from a bolus isotope infusion and a single scan - this has been the goal in developing the described technique. For scanning other organs, further studies will be required to learn whether adaptations can be more informative than present commonly used techniques.

Less expensive digital discs for mass storage are becoming available. The software which has been in use is now being modified for a fixed head disc operating system. Upon completion, scans of a few milliseconds duration can be secured from Anger type cameras. Permanent recording of the data can be handled after scanning is accomplished. Raw data can be input to the computer and a few seconds later, photography will record the computer output. Other time consuming steps will no longer be required in which additional paper tape had to be punched and read back into the computer.

ACKNOWLEDGEMENT

Supported in part by Grant HE 07628 from the National Heart and Lung Institute National Institutes of Health.

REFERENCES

1. R. O. Smith, W. D. Love, P. H. Lehan and H. K. Hellems, "Delayed Coronary Blood Flow Detected by Computer Analysis of Serial Scans", to be published in Am. Heart J.
2. W. D. Love and G. E. Burch, "Differences in the Rate of Rb^{86} Uptake by Several Regions of the Myocardium of Control Dogs and Dogs Receiving 1-Norepinephrine or Pitressin", J. Clinical Invest. 36(3), 479-484 (Mar. 1957).
3. W. D. Love and G. E. Burch, "Estimation of the Rates of Uptake of Rb^{86} by the Heart, Liver and Skeletal Muscle of Man with and without Cardiac Disease", Int. J. Appl. Radiat. Isot. 3, 207-216 (1958).
4. W. D. Love, R. O. Smith and P. E. Pulley, "Mapping Myocardial Mass and Regional Coronary Blood Flow by External Monitoring of ^{42}K or ^{86}Rb Clearance", J. Nucl. Med. 10(11), 702-707 (Nov. 1969).
5. R. O. Smith, "Computer Controlled, Intensity Modulated CRT Display of Isotope Scan Data", to be published in Proceedings of 2nd Symposium on the Sharing of Computer Programs and Technology in Nuclear Medicine.