Gamma Camera Image Recording on Video Cassette

John R. Turner* and Nigel J. Randall

St. Mary's Hospital, London, United Kingdom

The potential of the Anger scintillation camera cannot be fully realized unless data are recorded at the time of the patient study. Since many small nuclear medicine centers do not have access to a digital computer, there is a need for a simple and economical method to record and process data. A system has been developed that records static and dynamic scintigrams, without loss of image quality, where modified data from the camera are recorded on a modern video-cassette recorder in serial digital form. Results are presented showing the preservation of image quality at increasing count densities on a 127 × 127 digital matrix. Region-of-interest information may be extracted from the recorded data for simple indices or time-activity studies, alternatively, the stored digital data may be passed to a computer for further, more sophisticated processing.

The Anger scintillation camera (*1, 2*) is now an established imaging device that has found wide acceptance in clinical medicine *(3)*. As a result many small nuclear medicine departments have come into being but not all have the resources or indeed the need for a sophisticated computer system. The method to be described for recording image data is relevant to the requirements of such departments; it is very economical, easy to handle, and will provide simple on-site quantitation of data. Further, more sophisticated processing could be performed at a computer center at a later time if required, since all data are available, stored in a digital form. This approach is not proposed as an alternative to the existing mini-computer based systems already available. Rather, it is a simpler, more economical, albeit limited, approach to data handling that would add greatly to the potential clinical usefulness of a scintillation camera that would otherwise have no data recording or processing capabilities.

Up to the present time attempts to record image data without using a computer have involved the use of instrumentation *(4,5)*, video or digital tape recorders. The simplest approach has been to record the scintillation camera analog signals directly onto an instrumentation recorder. The method is restricted to low count rates (<10,000 cps) owing to the limited bandwidth of such recorders. It is also uneconomical on tape *(5)*. As a result, manufacturers have centered their attention on video or digital tape recorders. With the advent of fast, high resolution scintillation cameras such recording methods have been considered to be of limited use partly because of their count-rate limitations, but perhaps more importantly, because the replay must always be in real time. The former limitations are overcome in the digital recorder but the latter objection is not. Moreover, digital recorders are expensive items. The cost of such systems has been in the range of \$17,000– 24,000. Mini-computer data logging systems (employing floppy disks and a *small* central processing unit) are now available at around \$28,000 from, for example, Link Systems Ltd., U.K. These have limited processing facilities but do not have the real time limitations of the above. They are excellent for data storage and are a cheaper alternative to a full computer system, which may cost in excess of \$50,000.

At this center there was a requirement for an economical method of data logging, which would provide limited processing on site and would be mobile. We have developed a system that uses a Sony U-Matic ¾in. video-cassette recorder. The analog signals to the scintillation camera display are digitized before recording on tape, each scintillation being represented as a 16-bit word. Replay is immediate and can be by direct reconversion to the analog form or by computer manipulation and associated processing (e.g., image enhancement, functional images, deconvolution, transit-time analysis) at a later time. Image quality is maintained on a 127 × 127 digital matrix with moderate or slow dynamic studies. Time marks and physiological data may be recorded on two audio tracks. The system is at present being used for static and dynamic studies, analog region-of-interest (ROI) information being used on site to provide simple indices (ratio of regional count rate to whole organ count rate) and time-activity curves. The cost of the video-recorder and additional electronics has been less than \$3,000.

Choice of Recorder Type

There are several reasons for concentrating attention on on the video recorder as the basis for data logging. They are:

(1) Availability. These machines are very economical in comparison with "scientific" instruments such as instrumentation or digital recorders.

(2) Refinement. Continuing development has resulted

*Present address: Dept. of Medical Physics, St. Thomas' Hospital, Lambeth Palace Road, London, S.E.1. U.K.

For reprints contact: N.J. Randall, Dept. of Bio-Engineering, St. Mary's Hospital Medical School, Praed St., London W2. U.K.

in relative ease of use, e.g., totally electronic control giving foolproof cassette loading, tape handling, and fast access to any part of the tape.

(3) Transportability. Instruments available now are very suitable for transportation on a hospital cart or by hand.

(4) Ruggedness. In the U-matic-type instrument the tape transport system is highly developed yet protected by the automatic cassette-loading facility. Cassettes can stand much harsher treatment than floppy discs and this is a priority in a busy hospital department.

(5) Future uses. Remote electronic control of all mechanical operations leads to exciting possibilities with applications under the control of, for example, microprocessors.

With the above points in mind a method that can convert scintillation camera data to a suitable form for recording on video cassette should be desirable for small departments that do not have the resources to purchase a digital computer. Replay is still in real time but time compression is possible for slow dynamic studies (0.5–1.5 hr) by replaying only selected portions of the study (see (5) above), which is what happens with many computer systems.

Scintillation Camera/Video Recorder Interface

In order to maintain image quality at all times the system should not degrade the spatial resolution or field linearity of the original analog image *(6)*. To maintain those parameters the signal noise ratio (SNR) and the band-width of the video recorder should both be as high as possible; also the data transfer should be linear. We have measured the bandwidth and SNR on several modern video recorders and found that the bandwidth usually exceeds 3MHz but the SNR is very poor (~20dB). Video recorder manufacturers often quote SNRs >30dB; investigations have shown that these high values are very optimistic and result from frequency weighted noise measurements plus the averaged effects of signal dropout compensating circuitry. Our own investigations with a modern scintillation camera have shown that digitization of the analog signals *must* be performed if the image quality is to be maintained. In our method the analog signals are incorporated into a 16-bit digital word (Fig.1), which is written onto video tape using the normal frequency modulated (FM) carrier signal and helical scan system.

Figure 2 illustrates a block diagram of the record/replay system. The X and Y scintillation position signals are fed to the fast sample and hold (S/H) circuits and an analog-to-digital (A-D) conversion is accomplished on a nonvarying voltage level. This sequence is initiated by a pulse (prepulse) from the camera, which occurs just before the X and Y events. The X and Y digital forms are then serialized and preceded by a *higher amplitude,* word initiating pulse (WIP) for word decoding on playback. The last bit (to be used with a dual-isotope camera) is either logical 0 or 1; this information is provided from the camera's energy analyzers. On replay the output of a 15-bit shift register, loaded with the digital form of X and Y, plus the isotope bit may be converted to the analog form for on-site display of images

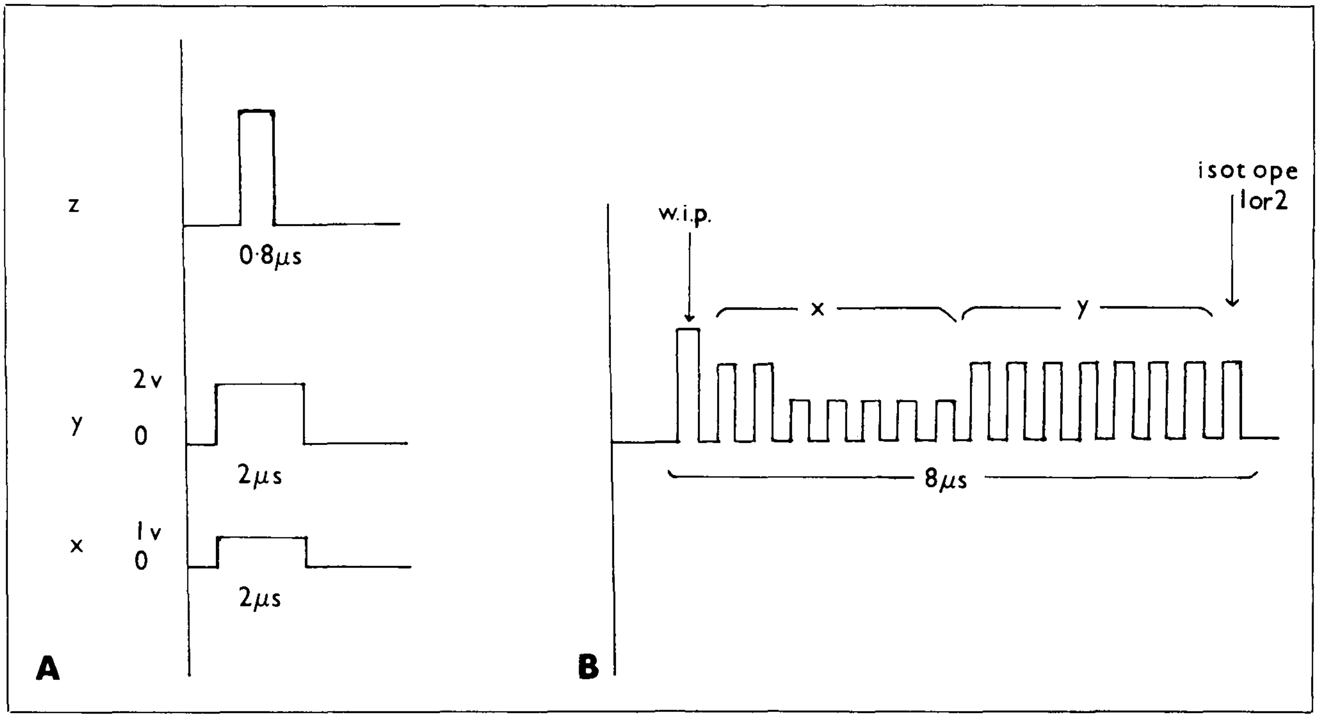


FIG. 1. (a) Analog signals from the scintillation camera. X and Y are bipolar, i.e., -2v to +2v. (B) Single digital word representing the data in A. Digital form of X and Y is coded in offset binary and is preceded by the word identification pulse (WIP).

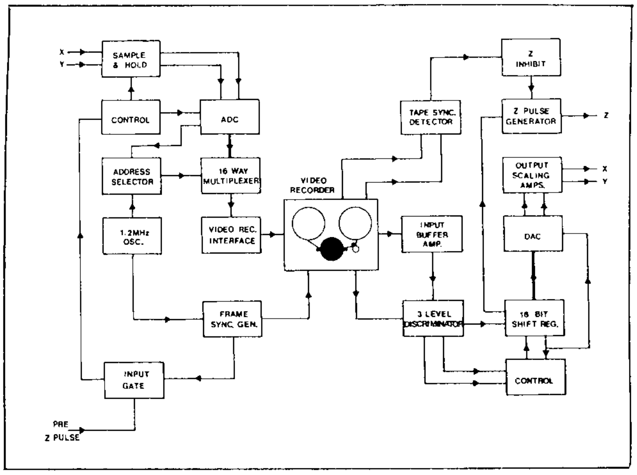


FIG. 2. Block diagram of record/replay system.

or kept in digital form for manipulation by a digital computer at a later time.

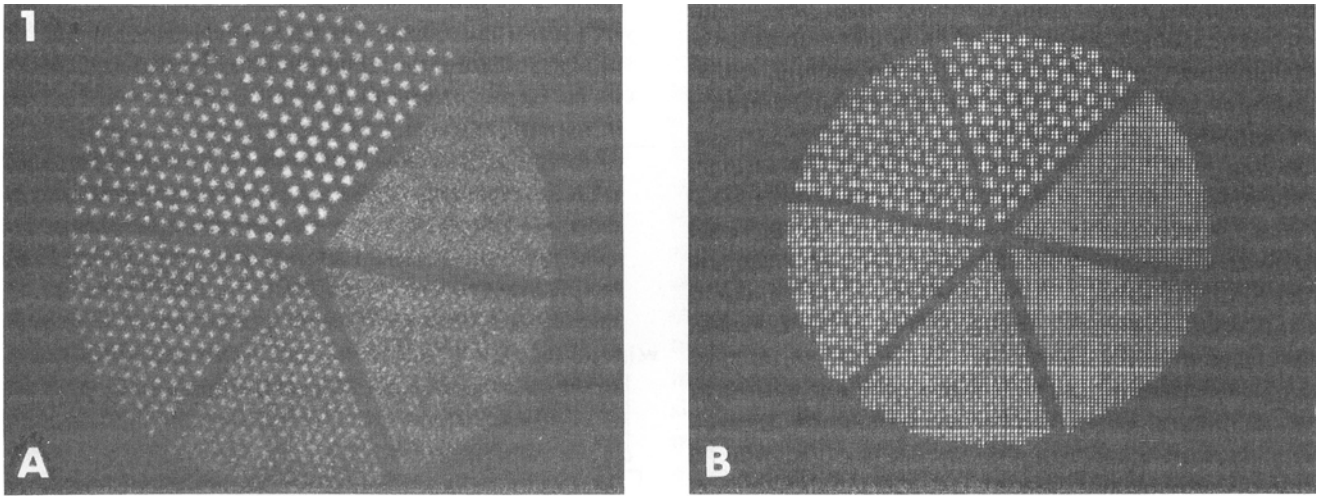
The format thus requires four voltage levels to be recorded:base line zero, data bit 0, data bit 1, and the WIP. This imposes a minimum SNR of only 12dB in the recorder to discriminate these levels, which is easily achieved. To prevent interference from a second scintillation occurring during the digital conversion and recording time of the first, an input gate blocks incoming data imposing a fixed electronic dead time. In the video method of recording the spinning heads are kept in synchronization with a reference pulse, this normally being the frame synchronization from a TV raster scan. In this application the reference pulse is derived from the master oscillator, which also provides the clock pulses for the serialization of the digital data.

On playback the signal is fed to a 3-level discriminator that extracts the following.

1. The clock pulses that will enable the data pulses to be loaded into the shift registers.
2. The data pulses representing X and Y by two 7-bit words plus the isotope identification bit.
3. The WIP, which after clearing the shift registers, enables them to receive the new data as read off from the tape. The data held are taken via two digital-to-analog (D-A) converters to the display with an accompanying regenerated Z pulse.

Results—Image and Count Rate Capability

Spatial Resolution and Field Uniformity: With the camera used in this investigation (Elscint Dymax L.F.) densitometric measurements of a line source of technetium-99m *(6)* resulted in an intrinsic spatial resolution of 6.0 (±0.2)mm FWHM in the original analog image; the re-



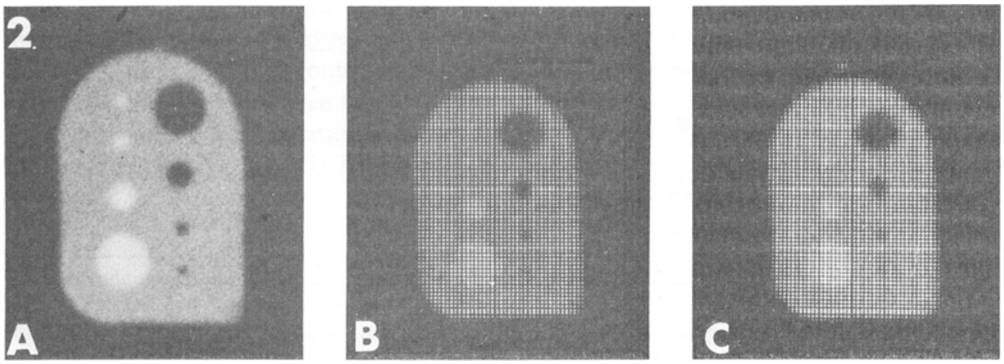


FIG. 3. Image replays. (1) the Anger phantom. (A) Original. (B) Replay. 0.5 × 10^6 counts in each case. (2) The Williams phantom. (A) Original. (B) and (C) Replays at low (7Kcps) and high (50Kcps) count rates respectively 1 × 10^6 total counts in each case.

played image was measured in a similar manner and found to have a FWHM of 10.2 (±0.2)mm. The degradation of spatial resolution in the present system is known to occur in the sample and hold (S/H) and A-D stages. With further attention and better A-D converters it should be possible to approach the limit of resolution dictated by the matrix, i.e., ~6.5mm. Figure 3 illustrates the recorded and replayed (quantized) images from the Anger pie phantom and the Williams phantom.

The bright lines seen on the replayed image are artifactual and arise from nonlinearities in the D-A converters. One, in particular, shows a gross nonlinear step that is repeated at every 2^4 interval. The D-A converters used are moderately fast settling devices (1 μs) but also very inexpensive and the system would benefit from the use of better converters in this respect. Better image quality was demonstrated when higher accuracy converters were used, but unfortunately these suffered from a slow conversion time. Continuing development of better and less expensive devices should soon overcome these problems.

Count Rate Capability: There are *two* main causes of count-loss in the system.

(1) The Poisson time-distribution of events. Since every analog display pulse is a direct result of a photon interaction in the scintillation crystal, the arrival time of these pulses is random. For N events occurring at a count-rate of μ (events/sec) the number of lost events is given by *(7)*:

$$n = N(1 - e^{-\mu\tau})$$

where τ is the dead time.

The present system requires 8 μsec for the complete conversion plus transfer to tape, thus the theory predicts a 10% loss of counts at 13 K counts/sec. A comparison of the count rate from the recording system with count rate measured by the scintillation camera ratemeter (derived from the pulse-height analyser) demonstrated agreement with the theory.

For most of the work of a small nuclear medicine department this increasing loss of counts at higher count rates does not impose severe limitations, for instance in our center, renal (DTPA and DMSA), biliary (HIDA), liver, bone, and brain studies all using Tc-99m do not involve count rates above 5 Kcps. However, electronic buffers which act as derandomizers between the camera and the recording device may be used, e.g., a first-in-first-out circuit (FIFO) *(9)*. The FIFO's purpose is to arrange the random pulses in a more regular sequence before passing them to the recording system. Computer simulation studies have shown that by using one or two FIFO stages the number of lost events may be kept well below 10% at high count rates *(9)*. Thus the 10% threshold may (if considered necessary) be extended to the limit in transferring the 16-bit word to tape, in our case ~130 K cps.

(2) Regular blanking time. Any video recording method will impose a short signal loss (~0.8 msec) at precise intervals of 20 msec for a 50 Hz head scan frequency, which occurs when the spinning recording heads are not in contact with the tape. With TV systems this is synchronized with the frame reset of the raster. We have imposed a fixed blanking period of 0.8 msec to cover this gap in signal recording by using the input gate; the result is a net *fixed* loss of 4% of original counts at *all* count rates. This loss does not have any serious detrimental effect on image quantitation as it remains a constant amount. Figure 4 demonstrates the relative loss of counts at increasing count rates. Initially the loss is constant due to the regular blanking time; then it increases exponentially due to the Poisson loss.

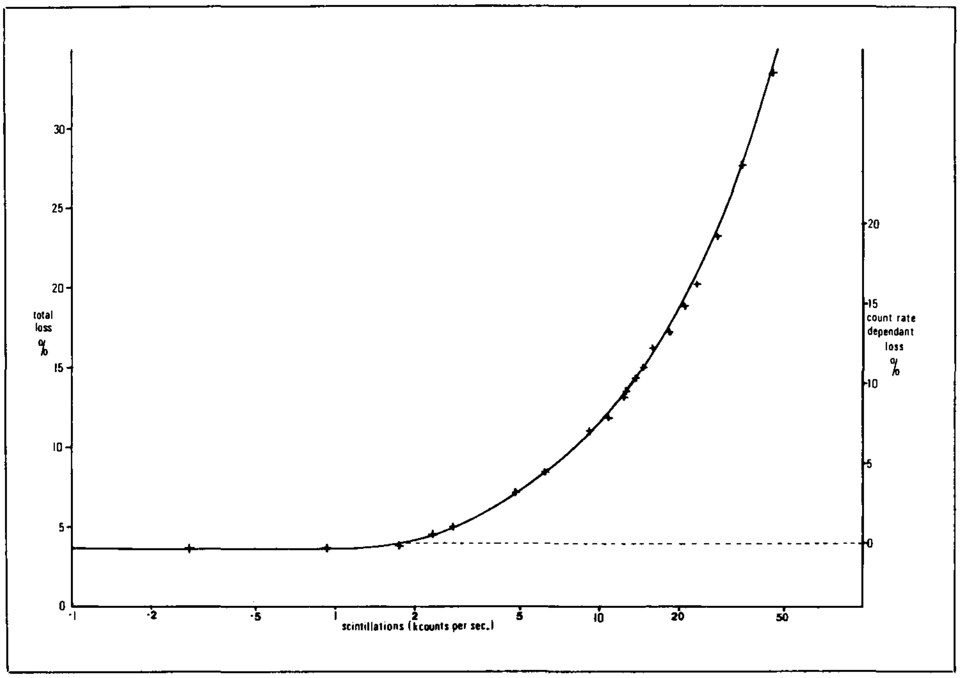


FIG. 4. Loss of counts as a result of imposed dead times. Curve is obtained from theory for a regular blanking time of 0.8msec every 20 msec (obvious at low count rates), and for an 8-μsec dead time during recording of each scintillation. Points represent measured count rate losses for this system without derandomizer.

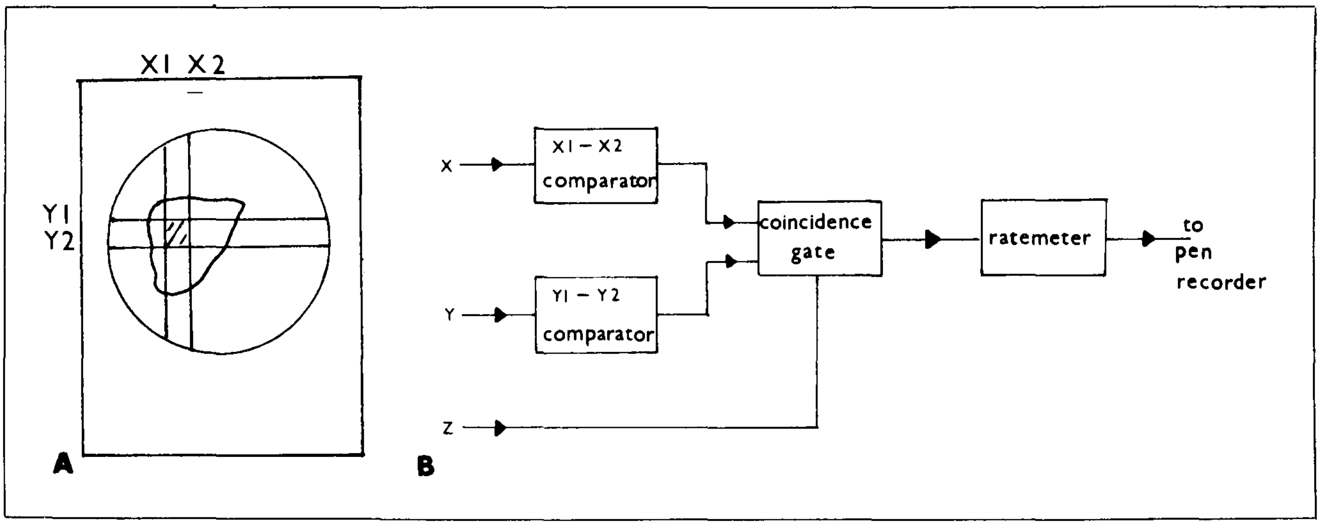


FIG. 5. (A) Display/monitoring oscilloscope. Lines demarcating region of interest (shaded) are generated during the 0.8 msec blanking period every 20 msec and appear superimposed over scintigraphic image. (B) Schematic diagram of simple analog region-of-interest analyzer.

Quantitation of Data

We have presented a very economical method for recording scintillation camera data without loss of image quality. Once in the stored digital form the data may be passed to a digital computer at a better equipped center for extraction of the required parameters. However, it is a distinct advantage to be able to obtain quantitative information on site, shortly after the patient study.

In order to obtain ROI information we used window comparators acting on the X and Y analog signals in order to discriminate an event within the chosen region bounded by X_1–X_2 and Y_1–Y_2 (Fig. 5). When coincidence occurs between the X and Y comparators a pulse is fed to the ratemeter. By taking advantage of the 0.8 msec blanking time every 20 msec during record and playback, lines demarcating the ROI rectangle can be drawn over the image by modulating the display oscilloscope with multiplexed ramp signals during this period.

The results of analyzing a Tc-99m DTPA study and a biliary Tc-99m HIDA study are shown in Fig. 6. For slow dynamic studies such as this we propose programming the video recorder to record at regular specified intervals in order to shorten the replay time; we have already performed many studies with a manual record interrupt to achieve this result. The modern U-matic recorder with its fully electronic controls is well adapted for this type of control. A comparison with a commercial floppy disk-based mini-computer showed that virtually identical time-activity curves were produced by the two systems.

The use of a derandomizer (many scintillation cameras now have built-in derandomizer circuits) will extend the count rate capability of the system to well above the count rate limit of the camera and can therefore allow the recording of fast dynamic studies. The U-matic recorder is equipped with two audio tracks that permit the recording of synchronous physiologic data and time marks. This is particularly useful for gated studies. It is therefore pos-

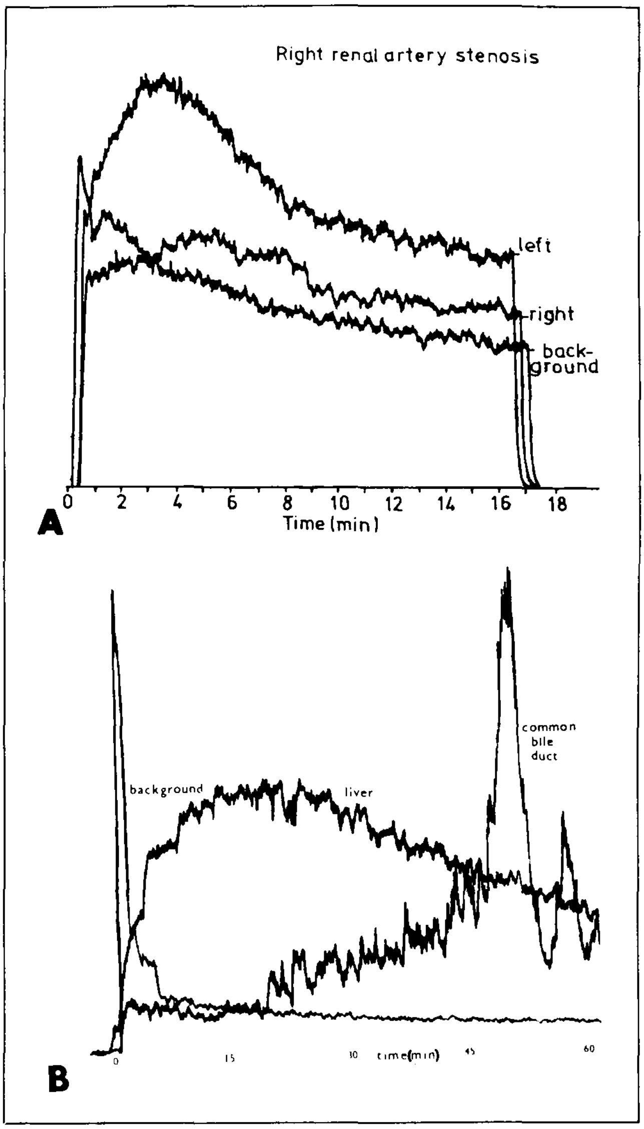


FIG. 6. Clinical studies using three regions of interest. (A) Tc-99m DTPA renogram; (B) Tc-99m HIDA biliary study.

sible to use this particularly mobile system for recording first pass gated cardiac studies.

Conclusions

A highly mobile, economical system has been described that enables the data from a gamma scintillation camera to be recorded without significant loss of image quality. Static and dynamic scintigrams are reproduced on a 127 × 127 digital matrix in real time and two tracks are available for physiologic data and time marks. The electronics additional to the video-recorder could be constructed in most medical physics departments at low cost; the total cost of the recorder plus additional electronics has been about $3,000.

Quantitation of data in ROI studies may be performed either on site using simple analog comparators and a rectangular ROI, or if required, the stored data may be passed to a computer for more sophisticated processing. Because of the Poisson nature of the incoming data from the scintillation camera, quantitation is limited to slow and moderate count rate studies; however, the addition of a derandomizing circuit may be expected to remove this limitation in which case the system may be used for fast (e.g., cardiac) dynamic studies.

We will be pleased to supply detailed circuit diagrams on request; please make this requirement clear when requesting reprints.

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