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A METHOD USING A DIGITAL COMPUTER FOR REDUCING RESPIRATORY ARTIFACT ON LIVER SCANS MADE WITH A CAMERA

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Respiratory motion is a major factor limiting the resolution obtainable on liver scans. It causes a disturbing "scalloped" appearance of the liver margins on rectilinear scans. This visual effect can be reduced by close spacing of the scan lines (*1*), but this does not recover the information that has been lost because of motion (*2*).

On liver scans made with a camera, respiratory motion degrades image quality by blurring margins. The better the intrinsic resolution of the camera, the more objectionable this blurring becomes. The blurring is increased if dyspnea is present, and frequently patients having liver scans are very ill and somewhat dyspneic.

An effective method is needed for reducing the respiratory motion artifact on liver scans made with a camera and recovering the information that has been lost because of this motion. One method is to have the patient hold his breath while the camera is collecting counts (*3,4*). We use this method routinely, but it increases the time and difficulty of the study and unfortunately cannot be used for very ill, dyspneic, or uncooperative patients. A method of compensating for the motion of the liver during respiration is needed. Various ways of accomplishing this have been suggested (*5,6*), all of which require some kind of detector external to the camera for determining the position of the liver. These methods face the technical problems of design, calibration, and maintenance of the required detector. Moreover, the detector must be properly applied to the patient, increasing the time and technical difficulty of the examination. To my knowledge none of these methods has yet been shown to be clinically useful on a routine basis.

A method is presented here for reducing the respiratory motion artifact which does not require a device separate from the camera for locating the liver but derives the necessary positional information from the counts collected by the camera itself in the course of a routine study. The study is subdivided into a large number of short time intervals. The intervals are so short that there is very little movement of the liver during any one of them, but enough counts can be collected during each interval to permit the position of the liver to be accurately determined in a manner to be described. It then becomes possible to reposition the counts to compensate for the changing position of the liver.

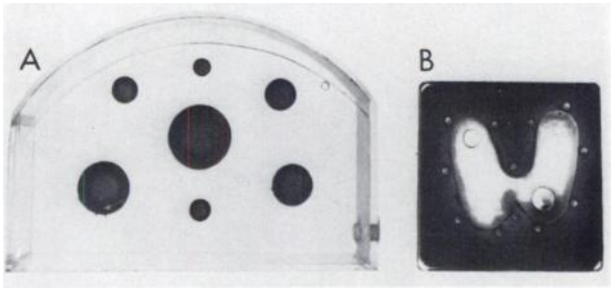

FIG. 1. Phantoms used in respiration studies. A is Liver phantom and B is Picker thyroid phantom.

MATERIALS AND METHODS

Studies were made with a liver phantom (Fig. 1A) and a Picker thyroid phantom (Fig. 1B). The liver phantom contained seven spheres 11, 13, 17, 21, 27, 34, and 43 mm in diameter, which became "cold" areas when radioactivity was introduced into the phantom. The thyroid phantom filled with radionuclide contained three "cold" areas and one "hot"

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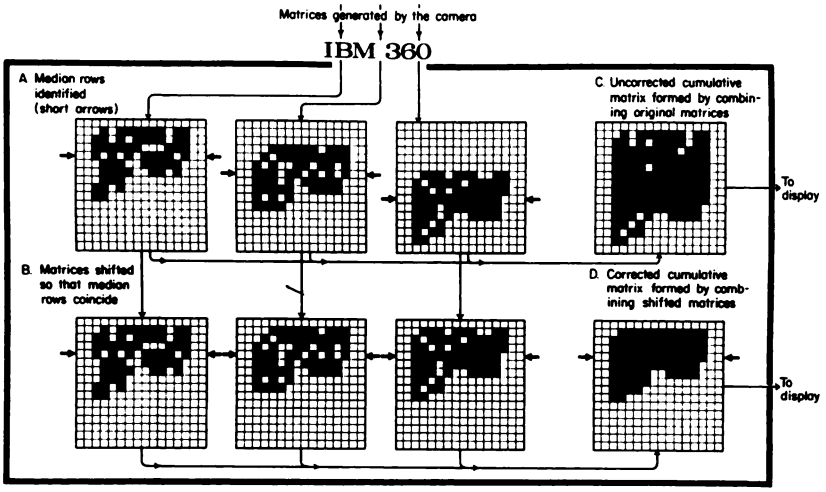

FIG. 2. Diagram of method used to correct for respiratory motion. Input into IBM 360 is counts from camera in form of series of matrices, one for each of short-time intervals into which study is subdivided. Output is two cumulative matrices, each containing all counts collected during study, one with motion uncorrected, other with motion correction applied. See text for definition of "median row" (short arrows).

area of varying sizes. The radionuclide used in both phantoms was ^{99m}Tc-pertechnetate.

The phantoms were scanned on a Nuclear-Chicago Pho/Gamma III camera, using a 4,000 parallel-hole low-energy collimator for the liver phantom studies and a pinhole collimator for the thyroid phantom studies. A counting rate of approximately 25,000 cps was obtained in all studies. Studies were made of both phantoms with no motion as well as with motion of varying amplitudes. Motion was produced by placing the phantom on a specially constructed respiratory motion simulator (*3*) capable of moving backwards and forwards in a manner analogous to respiratory motion. The amplitude and rate of motion could be varied over a wide range. For all studies with movement a rate of 5 cycles/min was used.

Each study was subdivided into 16 intervals of 0.6-sec duration, for a total collection time of 9.6 sec and a total collection of about 250,000 counts. During any interval the phantom moved 1/20th of a cycle. Counts collected during each interval were stored in a 64 × 64 matrix in the core memory of a PDP-12 computer. At the end of each interval the contents of the matrix were dumped onto a LINC tape mounted on the PDP-12, and later were transferred to a seven-track magnetic tape for processing by an IBM 360 Model 50 computer.

The manner in which a study was corrected for motion is illustrated in Fig. 2. First, the IBM 360 identified the "median row" of each of the 64 × 64 matrices. If counts in a matrix are summed row by row the median row is the first row for which the sum of counts exceeds half the total number of counts contained in the matrix. Second, each of the matrices was shifted by the computer so that its median row coincided with the median row of the first matrix. Third, the counts in corresponding locations of each matrix were summed to produce a cumulative 64 × 64 matrix containing all the counts for the study corrected for motion. An additional cumulative matrix with no correction for motion was produced for each study by combining the 16 matrices without shifting them. The uncorrected and corrected cumulative matrices were transferred to a seven-track magnetic tape, then to a LINC magnetic tape, and finally were displayed on the oscilloscope of the PDP-12 and photographed. The display used a 64-level gray scale with data blending.

RESULTS

Figure 3 shows the effectiveness of the method in correcting for motion on the studies made with the liver phantom. Figures 3A, 3C, and 3E were made without motion and with back-and-forth motions of 25 mm and 50 mm, respectively. Figures 3B, 3D, and 3F were made from the same data as Figs. 3A, 3C, and 3E with the method for motion correction applied. A considerable improvement in image qual-

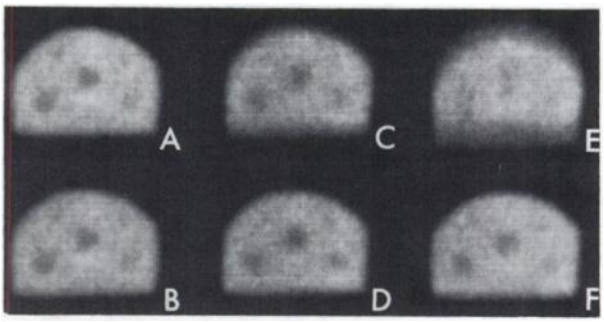

FIG. 3. Liver phantom studies. A. No motion of phantom, no correction for motion. B. Same data as in A, motion correction applied. C. 25-mm motion, uncorrected. D. Same data as in C, motion correction applied. E. 50-mm motion, uncorrected. F. Same data as in E, motion correction applied.

ity is apparent, with sharpening of the margins of the phantom, increased perceptibility of the "cold" areas, and elimination of the distortion caused by motion. Figure 4 similarly demonstrates the effect of the method on pinhole studies of the thyroid phantom. Figures 4A, 4C, and 4E were made without motion and with back-and-forth motions of 12 mm and 25 mm, respectively. Figures 4B, 4D, and 4F were made from the same data as Figs. 4A, 4C, and 4E with the method applied. Once again margins are seen to be much sharper, perceptibility of "cold" and "hot" areas is improved, and distortion from motion is eliminated.

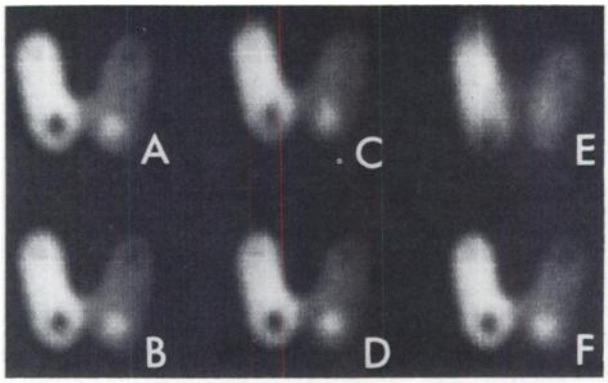

FIG. 4. Thyroid phantom studies. A. No motion of phantom, no correction for motion. B. Same data as in A, motion correction applied. C. 12-mm motion, uncorrected. D. Same data as in C, motion correction applied. E. 25-mm motion, uncorrected. F. Same data as in E, motion correction applied.

DISCUSSION

The method described uses only the counts collected by the camera to derive the positional information necessary to correct for motion. In clinical studies it would not require patient cooperation nor any alteration in the technique of a liver scan and would not be impaired by irregularities in the patient's respiration.

The theoretical aspects of this method have been investigated in detail and will be the subject of a subsequent publication. It can be shown that the "median row" is an accurate indicator of liver position. The effectiveness of the method is increased with decreased respiratory rate, increased depth of respiration, increased counting rate, or increased detector resolution. Enlarging the computer matrix increases its spatial resolution, but since this can never exceed the resolution of the detector, beyond a certain point any increase in matrix size results in negligible gain in resolution while entailing a considerable increase in the costs of equipment and processing. It has been shown (7) that for the equipment presently available the optimum matrix size for recording static images is 128 × 128. Either lengthening or shortening the time interval tends to impair the accuracy with which the liver can be localized. Lengthening the interval increases liver motion during each interval while shortening the interval decreases the number of counts collected during each interval, and thus increases the statistical error in determining liver position. The optimum time interval varies depending on the counting rate and respiratory rate. On theoretical grounds it appears that 0.25 sec would be a good compromise for clinical studies. For a respiratory rate of 12 breaths/min a 0.25-sec interval represents 1/20th of a cycle which is the fraction of a cycle covered by each interval in the phantom studies.

It can be demonstrated theoretically that with a counting rate of 600,000 cpm and storing counts for 0.25-sec intervals in a 128 × 128 matrix, the method should produce about a five-fold reduction in respiratory artifact for normal patients breathing quietly with greater reductions possible for dyspneic patients. With a counting rate of 400,000 cpm about a four-fold improvement should be obtained.

A comparison of the thyroid phantom studies with the liver phantom studies demonstrates the increased effectiveness of this method when detector resolution is improved. A greater improvement is noticeable between Figs. 4C and 4D, made with only 12-mm movement of the thyroid phantom but with the high-resolution pinhole collimator, than is noticeable between Figs. 3C and 3D made with 25-mm movement of the liver phantom but with a lower-resolution, multiple-hole collimator.

Clinical application of this method will place much more stringent requirements on the interface than the phantom studies cited here. The interface should meet the following specifications:

1. It should use a 128 × 128 matrix for storing counts although a 64 × 64 matrix might be used if a slight loss in resolution is accepted.

2. Either the dump time for a single matrix should be under 0.25 sec or the interface should contain a double core memory the size of two matrices and be capable of alternately storing into and dumping from each region. Otherwise, since dump time is deadtime with respect to count collection, clinical studies will be prolonged excessively.

3. The associated tape drive should be able to store at least 500 matrices on a single magnetic tape. With 0.25-sec intervals, 1 min of counting will generate 240 matrices, and the tape must be able to accommodate such vast amounts of data.

We are currently constructing an interface that will collect counts in a 128 × 128 matrix, will dump

an entire matrix in 0.14 sec, and will store about 1,500 matrices on a single tape.

While initially all processing will be done on an IBM 360 computer, we are seeking a cheaper and more convenient device for doing this. Processing on a medium-size computer appears feasible and will be attempted. Other computer configurations, such as task-specific computer module assemblies, are worthy of consideration. In addition, we are investigating analog techniques which will permit the method to be applied very conveniently and inexpensively if they are successful.

SUMMARY

A method has been devised for significantly reducing the respiratory motion artifact in liver images made with a gamma camera and recovering the information that has been lost by this motion. Other than data processing equipment, no ancillary devices are required. The method does not require patient cooperation, does not modify the technique of the study, and is not impaired by irregularities in the patient's respiration. The method uses a medium size computer as an interface and a large computer for processing. The effectiveness of the method has been demonstrated in phantom studies. A new interface is being constructed which should permit the successful application of this method to clinical material.

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REFERENCES

1. ATKINS HL, HAUSER W, RICHARDS P: Reduction of respiratory artifact in scintillation scanning. *Amer J Roentgen* 104: 682–685, 1968

2. QUINN JL: Editorial note. In *Year Book of Nuclear Medicine,* vol 4, Chicago, Year Book Medical Publishers, 1969, p 44

3. GOTTSCHALK A, HARPER PV, JIMINEZ FF, et al: Quantification of the respiratory motion artifact in radioisotope scanning with the rectilinear focused collimator scanner and the gamma scintillation camera. *J Nucl Med* 7: 243–251, 1966

4. GOTTSCHALK A, BECK RN, HARPER PV, et al: Imaging technetium-99m compounds with the gamma scintillation camera, and the use of this system to avoid the distortion of respiratory motion in radioisotope scintiphotography. In *Radioaktive Isotope in Klinik und Forschung,* vol 7, Munich, Urban & Schwarzenberg, 1967, pp 94–102

5. SMITH EM, SMOAK WM, MIBELLI J, et al: Use of physiologic triggers for dynamic and static Anger camera studies. *J Nucl Med* 10: 436–437, 1969

6. POTCHEN EJ: Comment in final panel discussion. In *Proceedings of the Symposium on the Role of Semiconductor Detectors in Nuclear Medicine,* Hoffer P, Beck RN, Gottschalk A, eds, New York, Society of Nuclear Medicine, 1971

7. ERICKSON JJ, BRILL AB: Quantitative approach to determination of minimum digital resolution in digitized camera images. *J Nucl Med* 11: 317, 1970