Computer Acquisition of Nuclear Medicine Images

Mark T. Madsen

Department of Radiology, University of Iowa Hospitals and Clinics, Iowa City, Iowa

Objective: The purpose of this article is to provide a comprehensive description of computer acquisition modes in nuclear medicine.
Methods: The paper discusses each acquisition mode in detail, explaining when use of each mode is justified.
Results: The effects of each acquisition mode on the resulting images are discussed.
Conclusion: Knowledge of acquisition modes and competence in acquiring digital images is a vital skill for nuclear medicine technologists.
Key Words: Computers, nuclear medicine studies, digital images.

J Nucl Med Technol 1994; 22:3–11

This is the first article in a four-part series on computers in nuclear medicine. Upon completion, the technologist should be able to list the types of acquisition modes for nuclear medicine studies, choose the best acquisition mode for any given study, and perform studies using these acquisition modes.

Most nuclear medicine imaging systems present their information as digital images. A digital image is stored in the computer as an array or matrix of count values and is displayed by assigning a gray or color scale that depends on the number of counts in each element. Typically (although not exclusively), the arrays are square matrices that have dimensions that range from 32 × 32 up to 1024 × 1024, although most nuclear medicine images have dimensions of either 64 × 64, 128 × 128 or 256 × 256 (*1,2*).

Each matrix element (commonly referred to as a pixel) is a location in computer memory. A 64 × 64 matrix has 4096 pixels, while a 128 × 128 matrix is four times larger (16,384 pixels), and a 256 × 256 matrix is 16 times larger (65,536 pixels). The number of counts which can be stored in a pixel depends on how many bits are allocated. Because of the way computers are designed, it is most convenient to assign either 8 bits (1 byte) or 16 bits (2 bytes) to each pixel. A 1-byte pixel requires half the storage space of a 2-byte pixel but has a limited dynamic range. The maximum number of counts that can be stored in a 1-byte pixel is 255 ($2^8 - 1$). In many instances, the likelihood of acquiring more than 255 counts is small, especially if the physical dimensions of the pixel are small (i.e., either a large matrix or a high zoom factor). However, there are circumstances where 255 counts per pixel is too limiting. The maximum number of counts which can be stored in a 2-byte pixel is 65,535 ($2^{16} - 1$).

For reprints contact: Mark T. Madsen, PhD, Dept. of Radiology, University of Iowa Hospitals and Clinics, 200 Hawkins Drive, Iowa City, IA 52242.

IMAGE FORMATION

When a gamma ray is absorbed by the sodium iodide crystal, a visible light flash known as a scintillation is created. This scintillation is sampled by an array of photomultiplier tubes that create a set of electronic pulses: an x and y position signal that corresponds to the location of the scintillation on the crystal, and an energy signal whose pulse height is proportional to the energy absorbed in the interaction. If the energy signal falls within a selected energy window, the x and y signals are digitized by a pair of analog-to-digital convertors. This position information can be stored in two ways: (1) as a sequential list of position entries referred to as list mode, or (2) by incrementing a matrix element (pixel) corresponding to the location of the position, known as frame or matrix mode (see Fig. 1).

Images in both frame mode and list mode can be acquired in a magnification or zoom mode. In zoom mode, the sampling is increased by digitizing over a smaller range of the position signals. This increases the sampling frequency (decreases the pixel size) and also decreases the field of view of the computer image. Typical zoom factors range from 1 to 4 in steps of about 0.25.

Frame Mode

Frame mode is the most common mode of image acquisition for nuclear medicine studies. Static, dynamic, gated, whole-body, and single-photon emission computed tomography (SPECT) studies are acquired in frame mode. With frame mode, a matrix of computer locations is cleared in memory prior to the start of acquisition. For each detected event, the appropriate matrix element is incremented. This

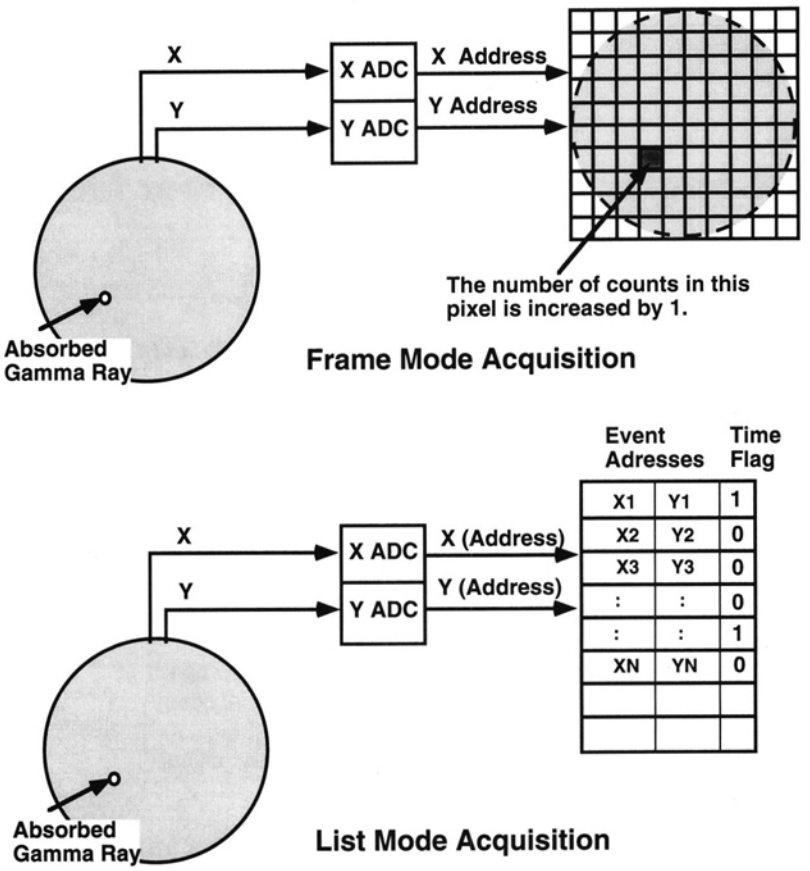


FIG. 1. Computer acquisition modes. Digitized signals from the gamma camera can be stored in list mode or can point to a pixel which is incremented, frame mode.

continues until a preselected time interval or total count value is reached. The memory or storage space required for a frame mode acquisition is determined solely by the matrix size and the number of frames acquired.

List Mode

List mode acquisition is less commonly used and some nuclear medicine computer systems no longer offer it as an option. Since information collected in list mode is a series of x and y locations, it cannot be viewed directly. It must be reconstructed into image matrices in the manner described above. The computer goes through the list of locations and increments a matrix element corresponding to that set of coordinates. The advantage of list mode studies is that the collected data can be framed in a variety of ways. One can alter the matrix size and the frame rate to suit a wide range of purposes. Physiologic gate signals can also be included in the list, allowing the reconfiguration of image data based on this information. The big disadvantage of list mode studies is that they require large amounts of computer space since each event is recorded separately.

Dual Isotope Imaging

Most gamma cameras allow the simultaneous selection of multiple photopeaks for gamma rays of different energies. This information can be stored in several ways. In the single isotope mode, the information from all of the selected photopeaks is added together. This is how multi-energy gamma ray emitters such as indium-111 (^{111}In) or gallium-67 (^{67}Ga) are frequently recorded; the detected counts from all the energy windows are combined into a single image. In the dual isotope mode, a separate frame (or list) is reserved for each selected photopeak (Fig. 2). This yields two distinct images that are spatially registered with one another, each of which has unique information.

One application of dual isotope imaging is technetium-99m (^{99m}Tc) MDP bone scanning in conjunction with ^{111}In white cell labeling for localizing sites of infection. Dual isotope image acquisition is currently available for all types of nuclear medicine imaging including SPECT and whole-body studies. All nuclear medicine computer systems permit dual isotope imaging and some can acquire image data from up to four separate energy windows.

SAMPLING

As previously described, the surface area of the gamma camera crystal is mapped into a matrix of computer elements (pixels). The physical size of a pixel is found by dividing the field of view by the matrix size, as shown in Figure 3. For example, if a 380 × 380-mm field is coincident with a 128 × 128 matrix, each pixel corresponds to a 3 × 3-mm area. If the matrix is changed to 64 × 64, the pixel size is increased to 6 × 6 mm. If an acquisition zoom factor of 2 is used, the field

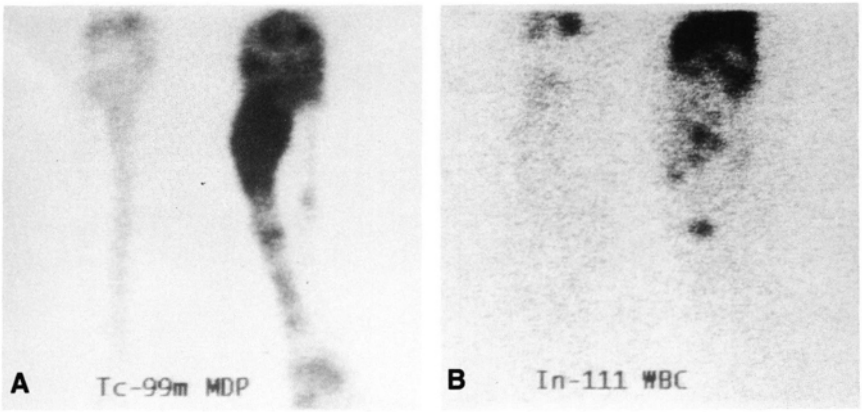


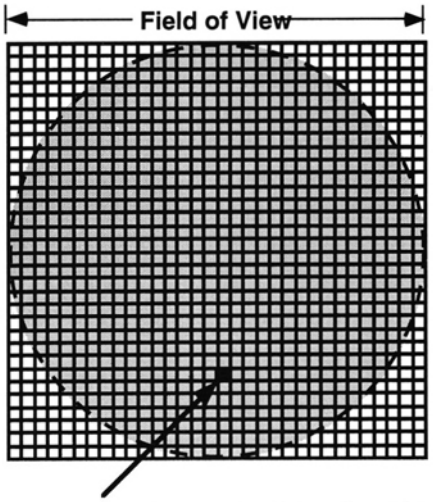
FIG. 2. Dual isotope imaging. Two separate images are acquired simultaneously with perfect spatial registration for each energy window. (A) ^{99m}Tc-MDP bone image; (B) ^{111}In white blood cell image.

of view will be decreased to 200 × 200 mm and the pixel size will now again be 3 × 3 mm. Since the spatial sampling of the image is determined by pixel size, increasing the acquisition zoom is one way to achieve sufficient sampling without using the additional storage space required for a larger matrix. However, this will only work for imaging smaller organs which can fit inside the reduced field of view.

By its very nature, a digital image is made up of discrete samples. The sampling interval is determined by the pixel size, which, as discussed above, depends on the matrix size and the image zoom factor. What should the pixel size be? The factors that determine the answer to that question are the following.

1. The spatial resolution of the imaging system.
2. The smallest object of interest in the image.
3. The time it takes to perform any processing steps.
4. The amount of storage and archival space available.

Factors 1 and 2 are related. Regardless of how small the object actually is, its image will be at least as large as the system response or spread function. If we are concerned about the smallest objects that can be seen on the image, then the limiting factor is related to the full-width-at-half-maximum (FWHM) of the spread function. In order to critically sample at this limit, the pixel size should be smaller than FWHM/3 (*3*). With a high resolution collimator, the FWHM for a source at 10 cm is about 9 mm. This implies that the pixel size should be no larger than 3 mm. If the gamma camera field of view is 380 mm, this would require a 128 × 128 matrix (3 mm pixel) or a 64 × 64 matrix with a zoom factor of at least 2.



Pixel Size (mm) = Field of View(mm)/ (# of Pixels)

FIG. 3. Pixel size is determined by matrix dimensions and size of field of view.

If the pixel size is coarser than this, the spatial resolution of the system is compromised and information is lost. This is illustrated in Figure 4, which shows images of a resolution phantom collected at several different pixel sizes with the total image count held constant. Note that successively larger hole patterns become obscured as the size of the pixel approaches the size of the holes in the pattern.

What are the disadvantages of oversampling, i.e., going to an even smaller pixel size? There are really only two disadvantages associated with oversampling, and these are both a consequence of using a larger matrix (Factors 3 and 4 listed above). Larger matrices require more storage space and processing time. For example, a 256 × 256 image takes up 4 times more space than does a 128 × 128 image, and there will be four times as many mathematical operations to perform. It should be noted, that there is *no* degradation in image quality associated with oversampling. Confusion seems to exist on this point because the statistical precision on a per-pixel basis gets worse as the size of the pixel decreases. While it is true that the counts per pixel decrease as the pixel size decreases, the count density (counts/cm^2) remains constant. This is also illustrated in Figure 4. There is a clear improvement in the quality of the image as it goes from being undersampled to critically sampled, with no discernible loss of quality as the pixel size is decreased beyond the critical sampling size of the smallest hole pattern.

INFORMATION DENSITY

Nuclear medicine images have statistical fluctuations which are a fundamental part of the emission and detection of gamma rays. The statistical precision of an image depends

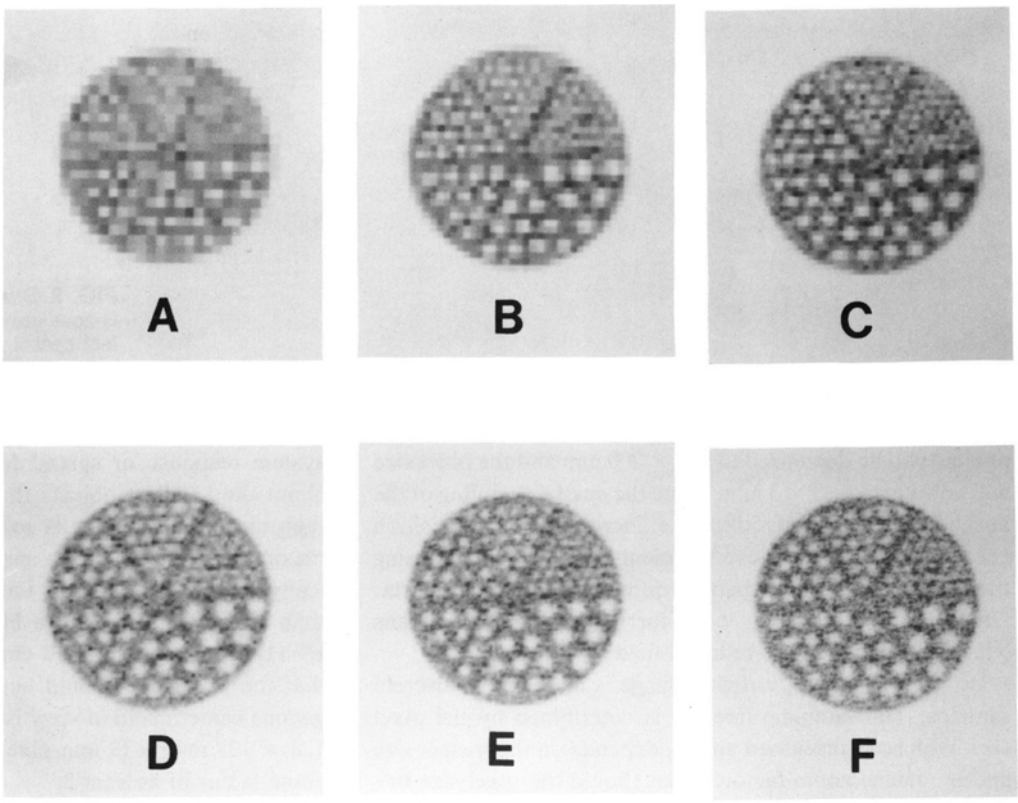


FIG. 4. Image quality as a function of pixel size: (A) 6.6 mm/pixel, (B) 4.8 mm/pixel, (C) 3.3 mm/pixel, (D) 2.4 mm/pixel, (E) 1.6 mm/pixel, (F) 1.2 mm/pixel. Sizes of holes in pattern are 16 mm, 12.7 mm, 11.1 mm, 9.5 mm, 7.9 mm and 6.4 mm, respectively.

on its count density, so a reasonable question to ask is how many counts should be acquired in an image. Although more counts are always better, there are constraints in the amount of activity a patient may be given and the length of imaging time. A better question may be: What information can we expect to perceive at a given count density? This depends on the size of the smallest region in the image you are trying to perceive and its apparent contrast to the surrounding background. Large objects and those with high contrast are very apparent even at low count density, while small, low-contrast objects are difficult to separate from the random noise. Simplified models of this imaging problem have been investigated for circular objects in a uniform background (*4*). From these one can estimate the count density (n) needed to perceive an area of the image of diameter (d) and image contrast (C), where contrast is defined as (object count density − background count density)/background count density, and k is the signal-to-noise ratio necessary to discriminate between random count fluctuations and true information.

$$n > k^2/C^2d^2 \qquad \text{Eq. 1}$$

Generally, k is in the range of 3–5 for digital images, and for illustration purposes, we will assume that a reasonable value for k^2 is 15.

So, suppose there is a circular region in the image which has a diameter of 1 cm and a contrast, with respect to its uniform background, of 0.1. The count density needed to perceive this must be greater than $15/(0.1)^2(1)^2 = 1500$ counts/cm^2. If the count density is significantly less than this, it will not be possible to reliably identify the object area from the random statistical fluctuations that are fundamental to radionuclide imaging. This is illustrated in Figure 5, which shows images of a hole pattern with 10% contrast acquired at varying count densities.

How does spatial resolution enter into this equation? It does so primarily through the contrast term. If the object under consideration has a true size that is about the same

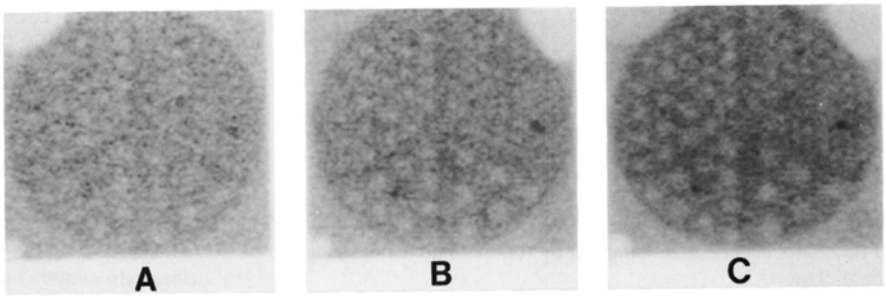


FIG. 5. Detection of information as a function of count density. (A) 625 counts/cm^2, (B) 1250 counts/cm^2, (C) 2500 counts/cm^2. Hole pattern has same dimensions as given in Figure 4, but contrast between holes and surrounding area has been set to 10%.

magnitude as the system's FWHM, then an improvement in spatial resolution significantly increases the image contrast; this makes the object more readily detectable at a lower count density. Therefore, efforts to improve spatial resolution without sacrificing sensitivity, such as moving the collimator closer to the patient, can greatly increase the ability to detect information in the image. Often gains in spatial resolution require losses in sensitivity, such as when a higher resolution collimator is used. Because of the large gains in contrast that are possible, it is often still advantageous to use high resolution collimators even with a significant decrease in counts. This will usually be the case in imaging studies where the available count rate is high (e.g., >1000 counts/sec). However, when the count rate is very low, the collimator choice should favor sensitivity over resolution.

Static Studies

Static studies are a collection of one or more spot views. The images are collected in frame mode for either a preselected time or total count level. Usually sufficient counts are collected to warrant the use of high resolution collimation and a small pixel size (≤3 mm/pixel) (*5*). With a large field-of-view gamma camera (≥500 mm) a 256 × 256 matrix is required to obtain a pixel of that dimension. Since spot views often have high count densities, the word mode (16-bit pixels) is used. When the acquisition of an individual frame is completed, the computer pauses to give the technologist an opportunity to reposition the patient and change the collection parameters (if necessary).

Whole-Body Imaging

There are several nuclear medicine studies for which the whole-body distribution of a radiopharmaceutical contains important information. One way of obtaining this information is to acquire a series of individual static views that covers the entire body. This can be time-consuming, and it is often difficult to view the whole body as a set of disjointed images. With whole-body imaging, the image of the entire body is incorporated into a single frame. The image is acquired by the relative motion of the patient past the detector with synchronized collection into a large array, typically 256 or 512 × 1024. The acquisition time (and count density) is determined by the scan speed, which is usually set to about 10–15 cm/min for ^{99m}Tc bone scans. This results in an acquisition time of about 15 min. Dual detector whole-body systems, consisting of two large, opposed gamma cameras, allow the simultaneous acquisition of anterior and posterior views (Fig. 6). Many of the currently available systems have autocontouring features which keep the anterior detector in close proximity to the patient throughout the scan.

Dynamic Studies

In many nuclear medicine studies, it is necessary to follow the uptake and clearance of a radiotracer over an extended period. This is often accomplished by acquiring a sequence of images, referred to as a dynamic study (*6*). Each frame in the sequence is acquired for a fixed amount of time that has been preselected by the operator. The frame rate can be as fast as 50 frames/sec or longer than 1 frame/hr. The number of frames that can be acquired is limited by the available disk storage. The acquisition of the image data is buffered so that while one frame is being acquired, previous frames can be stored to disk. Thus, there is no dead time between frames. Dynamic studies can be acquired in phases. Within each phase, the frame rate is fixed. For example, one might acquire a renal study with the following phases: 60 frames at 1 frame/sec (phase 1), 24 frames at 1 frame/5 sec (phase 2) and 60 frames at 1 frame/20 sec (phase 3).

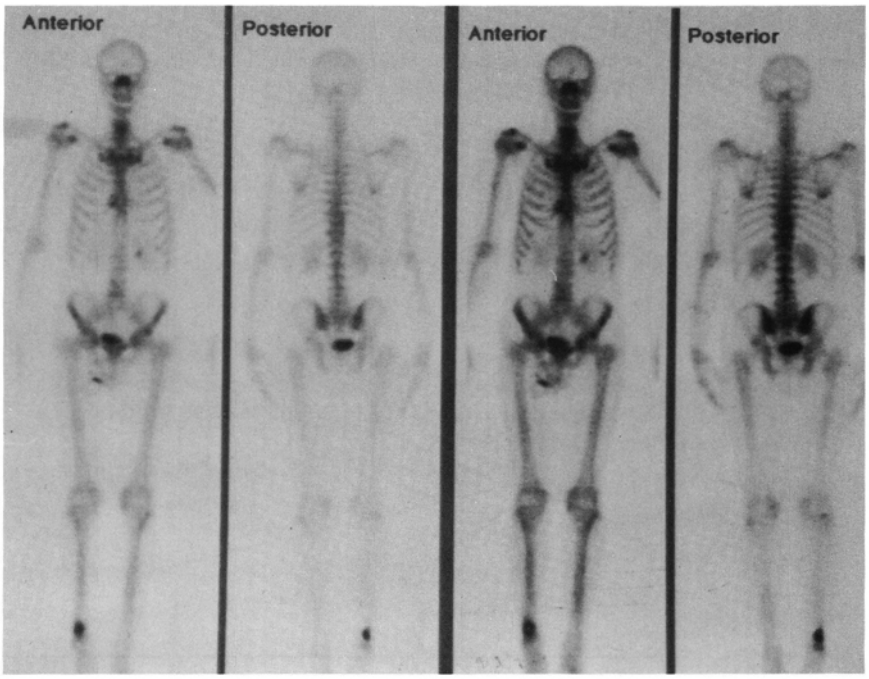


FIG. 6. Whole-body imaging. Two sets of anterior and posterior whole-body images acquired at different intensity settings.

What is the appropriate frame rate for a dynamic study? That depends on the kinetics of the radiotracer in the organ of interest. The time sampling should be no coarser than one-half of the shortest time interval under consideration. For example, suppose we wish to evaluate cardiac ejection fraction by a first-pass technique. In this case, the end systolic portion of the cardiac cycle has to be critically sampled. In a person with a normal heart rate, end systole persists for about 100 msec. Therefore, the frame duration should be no longer than 50 msec, which corresponds to a frame rate of 20 frames/sec.

The matrix size usually selected for dynamic studies is either 64 × 64 or 128 × 128. Although these selections may compromise the image quality somewhat because of spatial undersampling, they require much less storage space than a 256 × 256 matrix. In many cases the slight loss in spatial resolution is not a problem since the kinetics of the radiotracer may be the most important information gained from the study. Rapid dynamic studies are often acquired in byte mode in order to save storage space. Since the frame time in these studies is short, the pixel counts never approach 255, so there is little chance of count overflow.

Gated Acquisition

Red bloods cells can be tagged with ^{99m}Tc in order to measure heart function parameters such as ejection fraction. One way of doing this study is as a rapid dynamic acquisition during the first pass of an injected bolus (*7*). Although this is possible, there are several disadvantages to this approach, including limited count collection and time-consuming analysis. A dynamic study acquired after the radiotracer has equilibrated is not practical. Since the heart cycle is on the order of 1 sec, the frame rate would have to be on the order of 20 frames/sec and there would not be enough counts in each frame to adequately define the chambers.

The idea of gating the acquisition with the R-wave signal of the ECG signal was introduced in the mid-1970s (*8,9*). If we can assume that the heart moves with a regular and reproducible pattern during each beat, then it is possible to build up sufficient counts in the images by adding a large number of beats together. The images are acquired in the following manner (see Fig. 7). Prior to the beginning of the study, a number of frames (16–32) are reserved in image memory. When the first gate signal is received, the computer acquires counts into the first frame for a fixed time interval (usually 20–50 msec). At the end of this time, the information automatically is directed to the second frame for the same time interval. Each frame is acquired sequentially in this manner until a new gate signal is detected. At this time, the computer immediately begins acquiring into the first frame again and the sequence is repeated. This continues until sufficient counts are collected.

This method works well if the heart cycle is regular, but encounters problems when it is not. An early beat corrupts the sequence. On older computer systems, the system would stop acquiring when it sensed a significant change in the cardiac rhythm. However, image data were unavoidably collected during the first bad beat and could not be eliminated. On current systems, there is sufficient memory and processing speed to reject bad beats as they occur.

Another alternative to rejecting bad beats is to conduct a gated study in list mode. In this type of study, the R wave occurrences are recorded along with the count position data. After the study is acquired, an R interval histogram can be generated allowing the operator to select the desired beat lengths to analyze. The study is then framed into a gated format in the same manner as described above.

SPECT ACQUISITION

SPECT provides three-dimensional information about the distribution of radionuclides within a patient (*10*). This significantly improves the contrast and the anatomical localization of the radiotracer and therefore significantly increases the diagnostic utility of nuclear medicine imaging. A SPECT

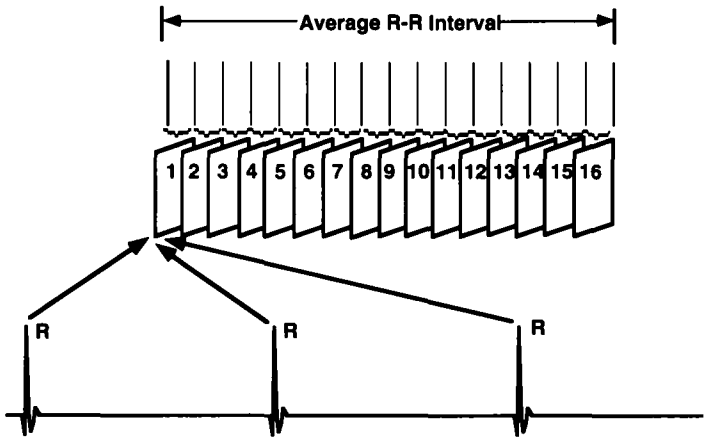


FIG. 7. Gated acquisition. Average interval between gate signals is divided into equal size time bins. Image data is repetitively acquired into these frames synchronized by gate signal.

study is a set of static images, often referred to as projections, collected at equal angular intervals from around the patient. The information in these projections is reconstructed into tomographic slices through a mathematical operation known as filtered backprojection. The reconstruction process relies on several basic premises, which are discussed below.

The first assumption is that the data collected represent "true" projections (i.e., the pixel counts are proportional to the sum of activity along the ray path defined by the collimator). This is not a good assumption because of the attenuation of body tissues. In order to compensate for this, information has to be acquired over 360° (see below) and additional mathematical corrections for attenuation must be made.

Another assumption is that the objects being imaged remain stationary throughout the study. Thus, patient motion must be minimized. A further assumption is that the center of rotation (COR) is accurately known. The SPECT detector(s) rotate about an imaginary line in space called the axis of rotation. In order to accurately reconstruct the tomographic slices, the location of where the axis of rotation projects onto the detector (the COR) must be accurately known. Deviations from the true COR must be less than 0.5 pixels to avoid a significant loss of spatial resolution. The COR is determined by acquiring a SPECT study of a line or point source according to the manufacturer's recommendations.

Another common assumption is that the detector has uniform sensitivity over its field of view. Nonuniformity in the camera's field of view results in a ring artifact in the reconstructed image that is concentric to the COR. Uniformity artifacts are most apparent when the nonuniformity is close to the center of the projection and when large, uniform distributions of activity are imaged. While the uniformity of gamma cameras was a major concern with the first SPECT systems, it is much less of a problem with current instrumentation.

The first SPECT systems consisted of a single gamma camera mounted to a gantry which could only rotate through a circular contour. Over the last 5 years, SPECT systems have evolved into sophisticated machines with two or more detectors. In addition to having a two- to threefold increase in sensitivity, the dedicated SPECT systems have intelligent gantries that can position the detectors close to the patient throughout the study, thereby improving spatial resolution.

The following parameter selections affect the acquisition of a SPECT study (*10,11*).

- Pixel size
- Arc selection
- Number of angular samples
- Acquisition time at each angle
- Rotation mode

The principles of sampling for SPECT projection images are similar to those discussed for static images: the pixel size should be smaller than the FWHM/3 of the final (reconstructed) image. Thus if the resolution of the reconstructed image is 15 mm, the pixel size should be no larger than 5 mm. If the pixel size is larger than this, there will be a perceptible loss of resolution, as shown in Figure 8.

As discussed above, there is no degradation in image quality associated with oversampling, but there is a penalty with respect to storage space and processing times. It should be noted that the spatial resolution available from the dedicated multi-detector SPECT systems can be below 10 mm for brain studies and in other studies where the detectors are in close proximity to the source. For these cases, a pixel size on the order of 3 mm/pixel is a necessity.

Since there are two sets of unknowns in SPECT imaging (radionuclide distribution and attenuation), SPECT studies

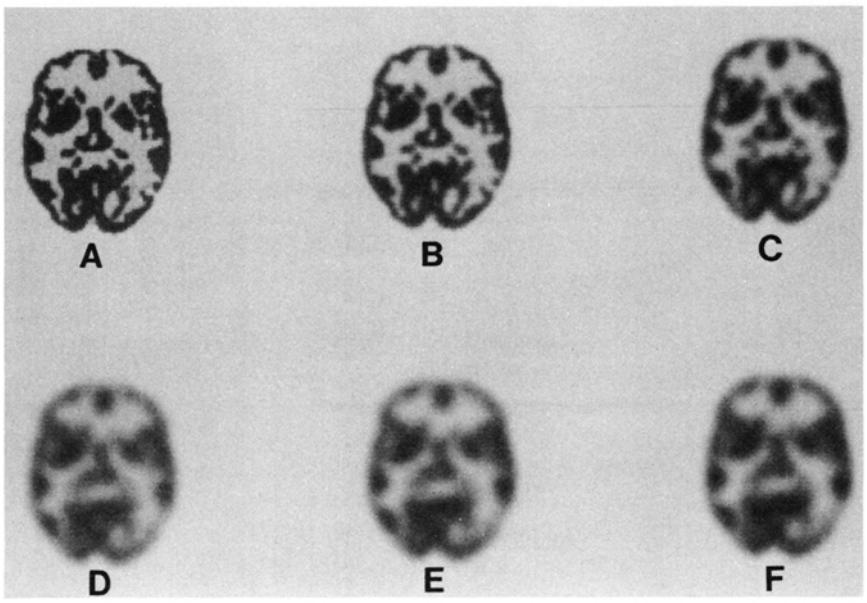


FIG. 8. SPECT image quality as a function of pixel size. (A) 2 mm/pixel, (B) 3 mm/pixel, (C) 4 mm/pixel, (D) 5 mm/pixel, (E) 6 mm/pixel, (F) 7 mm/pixel. As pixel size increases, it becomes a significant factor in resolution degradation.

are usually collected over 360°. However, if the source organ is eccentrically located (e.g., the heart), the data from some of these views may contain little useful information. In such cases, there can be a contrast improvement if the collection arc is restricted to less than a full 360°. In order to be free of reconstruction artifacts, the collection arc should be integral multiples of 180°, unless the software performs a weighted average of the additional views. For example, suppose 240° of data is collected. The additional 60° of data beyond 180° should be averaged (not summed) with its conjugate view.

Angular Samples

The number of angular samples or projection views required in a SPECT study depends not only on the expected resolution in the final reconstructed image, but also on the size and location of the organs of interest. This is because the sampling density decreases as the radial distance from the COR increases. The number of views required during a 360° rotation can be calculated using the following equation.

$$\text{Number of views} = 2\pi R/(\text{FWHM}), \qquad \text{Eq. 2}$$

where R is the radial distance from the COR that encompasses all objects of interest. For example, if the resolution of the reconstructed image is 15 mm and the organs of interest are all within a 15-cm radius from the COR, we would need $2\pi 15/(1.5) = 63$ angular samples over a 360° rotation.

Typically, the number of views is a factor of 32 (i.e., 64, 96, or 128) or is based on an integral angular increment (e.g., a 4° increment yields 90 views). If the number of views acquired is only moderately smaller than that required by Equation 2, little degradation is seen. However, when the number of views is much smaller than required, the image breaks apart, especially at the periphery, as demonstrated in Figure 9.

Acquisition Time

Although collecting high count density images is desirable, the length of time to collect each SPECT projection is determined primarily by patient comfort (the patient must lie motionless) and throughput needs. These factors constrain the total acquisition time to a range of about 10–30 min. The acquisition time for each frame is approximately equal to the total study time divided by the number of projection views. For example, if the total study is expected to be 15 min and there are 90 views, then the acquisition time at each view should be set to 10 seconds.

Rotation Mode

Most SPECT systems can acquire information in either a step-and-shoot (S&S) mode or a continuous rotation (CR) mode. The gantry rotates over an angular increment and then begins collecting data in the S&S mode, but collects and rotates simultaneously in the CR mode. The spatial resolution in the S&S mode is marginally better than of the CR mode, but the CR mode is more efficient since it has no dead time. The CR mode is also quieter and may be less stressful to some patients. However, in some SPECT systems, the CR mode only works for circular contours.

Collimator Selection and Patient Setup

Both good resolution and high count density are important for SPECT. If we assume that there is a fixed amount of time to acquire a SPECT study, should the collimator be chosen for more sensitivity to obtain high count density, or for higher spatial resolution? A number of studies have shown that, for studies where the available count rate is high (>2000 counts/sec), high resolution collimation is preferred

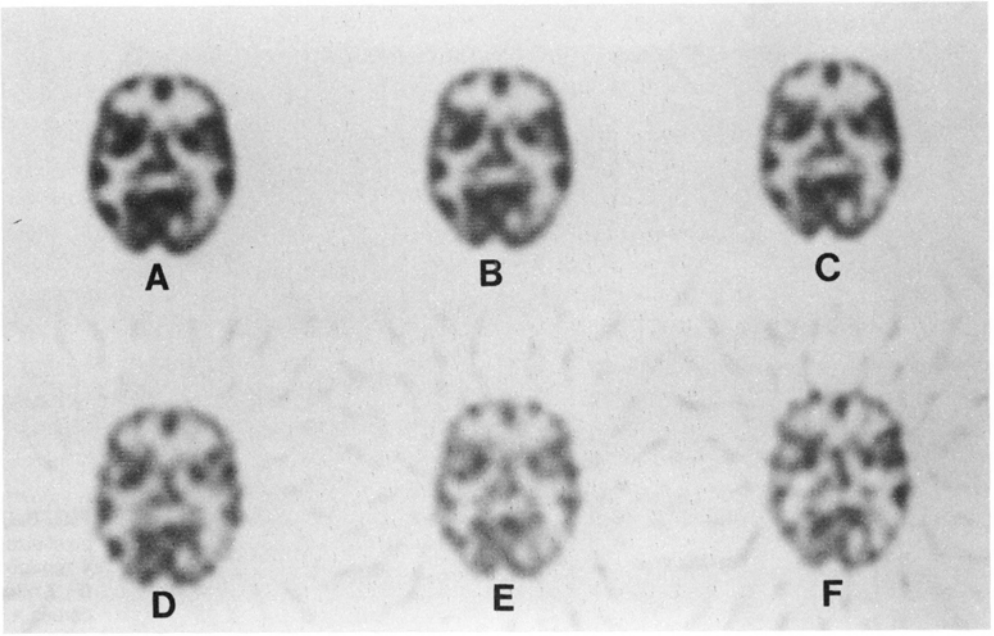


FIG. 9. SPECT image quality as a function of number of projection views. (A) 120 projections, (B) 90 projections, (C) 72 projections, (D) 60 projections, (E) 45 projections, (F) 40 projections.

(*12,13*). This is especially true for multi-detector SPECT systems.

However, the gain in resolution in going from a general purpose collimator to a high resolution collimator is about a 1-mm improvement in the FWHM. One can get an improvement of this magnitude without changing collimators (or losing any sensitivity) by moving the detectors 2 cm closer to the patient. The point of this observation is that patient setup is as important as collimator selection and every effort should be made to keep the collimators as close to the patient as possible throughout the acquisition. However, this needs to be done efficiently so as not to increase the time the patient is forced to lie still. This is being facilitated in many of the newer SPECT systems, which have automatic sensing devices for determining the optimal patient contour.

In organs such as the brain, fan beam collimation can be used to obtain improved sensitivity at a specified spatial resolution (*13*). Since the field of view decreases with distance from the collimator, patient setup is critical for fan beam SPECT in order to avoid losing information at the periphery. All the acquisition parameters discussed above are similar for fan beam SPECT, but a specialized reconstruction algorithm must be used.

Gated SPECT

The length of time each projection image is collected is long compared to the cardiac cycle. As a result, there is a significant loss of resolution when imaging the heart. As a way of "freezing" the heart motion and thereby improving spatial resolution, it is possible to collect SPECT projections as a series of gated studies (*14,15*). Gated SPECT studies have been used to improve the quality of myocardial perfusion studies as well as to obtain tomographic information about left ventricular function. At each stop in a gated SPECT acquisition, a gated study is acquired instead of a single projection view. Typically, the R to R interval is divided into 8 frames, so that the total information acquired is greater by a factor of 8 than for a nongated study. Therefore, these studies are generally acquired in 64 × 64 matrices and with an angular increment of about 6°.

Whole-Body SPECT

The axial field of view of a detector in a SPECT system falls somewhere in the range of 20 to 40 cm. There are instances when it would be preferable to have a SPECT study that encompassed a large portion of the body. This can be accomplished on some SPECT systems through a coordinated SPECT acquisition and table translation. Immediately following acquisition of all the projection views of the first SPECT study, the SPECT table is translated (under computer control) for the length of the axial field of view and a second SPECT study is commenced. This can be repeated several times in succession. After each of the individual SPECT studies is reconstructed, the images are reassembled so that the image field in the axial direction is extended.

CONCLUSION

In this paper, the various techniques associated with the computer acquisition of nuclear medicine images have been reviewed and the important imaging parameters have been examined. In the past, nuclear medicine images were generated by analog electronics and were recorded directly onto film as static or dynamic studies. Now, the computer has become an integral part of all nuclear medicine imaging systems, and all information is generated and presented as digital images. The acquisition of digital images for all types of studies requires the selection of the matrix size, magnification level, and acquisition times. These are determined by the spatial resolution, dimensions of the field of view, and the available count density. This allows acquisition and processing of a wide range of additional studies, including gated heart, whole-body, dual isotope, and SPECT studies.

REFERENCES

1. Weber DA. Computers in nuclear medicine: introductory concepts. *Semin Nucl Med* 1978;8:107–112.
2. Gelfand MJ, Thomas SR. *Effective use of computers in nuclear medicine*. New York: McGraw-Hill; 1988:1–30.
3. Sorenson JA, Phelps ME. *Physics in nuclear medicine*. 2nd ed. Philadelphia, PA: W.B. Saunders; 1987:401.
4. Rose A. *Vision: human and electronic*. New York: Plenum; 1973.
5. O'Connor MK, Brown ML, Hung JC, et al. The art of bone scintigraphy—technical aspects. *J Nucl Med* 1991;32:2332–2341.
6. Craddock TD, MacIntyre WJ. Camera-computer systems for rapid dynamic imaging studies. *Semin Nucl Med* 1977;7:323–336.
7. Berger HJ, Matthay RA, Pylik LM, et al. First-pass radionuclide assessment of right and left ventricular performance in patients with cardiac and pulmonary disease. *Semin Nucl Med* 1979;9:275–295.
8. Alpert NM, McKusick KA, Pohost GM, et al. Noninvasive nuclear kinecardiography. Concise communication. *J Nucl Med* 1974;15:1182–1184.
9. Strauss HW, McKusick KA, Boucher CA, et al. Of linens and laces—the eighth anniversary of the gated blood pool scan. *Semin Nucl Med* 1979; 9:296–309.
10. Phelps ME. Emission computed tomography. *Semin Nucl Med* 1977;7: 337–365.
11. Fahey FH, Harkness BA, Keyes JW, et al. Sensitivity, resolution and image quality with a multi-head SPECT camera. *J Nucl Med* 1992;33: 1859–1863.
12. Moore SC, Kouris K, Cullum I. Collimator design for single photon emission tomography. *Eur J Nucl Med* 1992;19:138–150.
13. English RJ, Brown SE. *Single-photon emission computed tomography: a primer*. 2nd ed. New York: Society of Nuclear Medicine; 1990:47–68.
14. Fischman AJ, Moore RH, Gill JB, et al. Gated blood pool tomography: a technology whose time has come. *Semin Nucl Med* 1989;19:13–21.
15. Morgan MG, Mannting F. Practical and diagnostic considerations for gated myocardial perfusion tomography using sestamibi. *J Nucl Med Technol* 1993;21:13–19.