The AAPM/RSNA Physics Tutorial for Residents

Physics of SPECT

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Single-photon emission computed tomography (SPECT) provides three-dimensional (3D) image information about the distribution of a radiopharmaceutical injected into the patient for diagnostic purposes. By combining conventional scintigraphic and computed tomographic methods, SPECT images present 3D functional information about the patient in more detail and higher contrast than found in planar scintigrams. A typical SPECT system consists of one or more scintillation cameras that acquire multiple two-dimensional planar projection images around the patient. The projection data are reconstructed into 3D images. The collimator of the scintillation camera has substantial effects on the spatial resolution and detection efficiency of the SPECT system. Physical factors such as photon attenuation and scatter affect the quantitative accuracy and quality of SPECT images, and various methods have been developed to compensate for these image-degrading effects. In myocardial SPECT, an important application of SPECT, recent use of attenuation compensation methods has provided images with reduced artifacts and distortions caused by the non-uniform attenuation in the chest region and by the diaphragmatic and breast attenuation. Attenuation-compensated myocardial SPECT images have the potential to improve clinical diagnosis by reducing the false-positive and false-negative detection of myocardial defects. In the future, further improvement in SPECT images will be realized from the continuous development of new radiopharmaceuticals for new clinical applications, instrumentation with high spatial resolution and detection efficiency, and image reconstruction algorithms and compensation methods that reduce the image-degrading effects of the collimator-detector, attenuation, and scatter.

INTRODUCTION

Single-photon emission computed tomography (SPECT) is a medical imaging modality that combines conventional scintigraphic and computed tomographic (CT) methods (1–4). Unlike x-ray CT, which uses radiation from an external x-ray source, SPECT uti-
lizes radiation, mostly gamma rays, from radio-
uclide-labeled pharmaceuticals (ie. radiophar-
maceuticals), which are administered to the pa-
tient and which become distributed in different
internal tissues or organs. SPECT differs from
positron emission tomography (PET) in the
types of radionuclides used and as a result of
the different imaging instrumentation used. PET
studies use radionuclides that emit positrons
with subsequent emission of two coincident
511-keV annihilation photons (eg. fluorine-18 or
carbon-11) and require imaging systems that are
capable of coincidence detection of the two
photons. SPECT studies use standard radionu-
clides (eg. technetium-99m or iodine-123) and
simpler imaging instrumentation. These stan-
dard radionuclides commonly emit gamma-ray
photons with energies that are much lower
than 511 keV. A typical example is Tc-99m,
which emits 140 keV photons. An exception is
the widely used myocardial agent, thallium-201,
whose decay to mercury-201 results in the
emission of characteristic mercury x rays with
an average energy of about 72 keV.

Both physical factors, such as attenuation
and scatter of gamma-ray photons in the pa-
tient, and instrumentation factors, such as de-
tection efficiency and spatial resolution of the
collimator-detector system, severely degrade
SPECT images (5,6). These factors result in high
image noise, poor resolution, low contrast, and
reconstruction artifacts and distortions. New
SPECT systems (3,7) and special collimator de-
signs (8) have provided increased detection effi-
ciency or improved spatial resolution. Quantita-
tive reconstruction methods that accurately
compensate for the image-degrading factors
have improved the quality and quantitative ac-
curacy of SPECT images, compared with what
could be achieved with conventional reconstruc-
tion methods (5). The advances in SPECT
instrumentation and image reconstruction
methods, combined with use of new radiophar-
maceuticals, have made SPECT one of the major
imaging modalities in clinical diagnosis.

This article represents only a brief review of
SPECT. A list of review articles is provided as
references. They also serve as sources of refer-
ces to specific research work in SPECT. This
article introduces the basic principles of SPECT.
Planar scintigraphy and techniques of image re-
construction from projections are briefly re-
viewed, since they are part of the basic com-
ponents of SPECT. The design concepts and re-
cent advances of SPECT instrumentation are
discussed, and factors that adversely affect
SPECT images are introduced. Quantitative re-
construction methods that compensate for
these image-degrading factors and that allow
production of high-quality SPECT images with
accurate quantification are described. This im-
provement in image quality (compared with
that achievable with conventional reconstruc-
tion methods) is demonstrated with phantom
and clinical images.

**CONVENTIONAL SCINTIGRAPHY**

Figure 1 shows components of a typical con-
ventional scintigraphy system. A given radioiso-
tope-labeled pharmaceutical is injected into the
patient. Depending on the biodistribution prop-
erties of the pharmaceutical, the radioactivity is
distributed in specific organs and normal or ab-
normal tissues. Gamma-ray photons emitted
from the radiopharmaceutical interact with
matter inside the patient’s body. Those photons
that exit the patient’s body and pass through
the holes of a collimator are detected by a posi-
tion-sensitive detector, typically a scintillation
(or Anger) camera mounted on a rotating gan-
try. A typical large field-of-view scintillation
camera consists of a 0.95-cm-thick, 40-cm-diam-
eter sodium iodide crystal and an array of pho-
Radionuclides Commonly Used in Scintigraphy and SPECT

<table>
<thead>
<tr>
<th>Radionuclide</th>
<th>Half-life</th>
<th>Energies of Primary Photons (keV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Technetium-99m</td>
<td>6 h</td>
<td>140</td>
</tr>
<tr>
<td>Thallium-201</td>
<td>73 h</td>
<td>~72</td>
</tr>
<tr>
<td>Gallium-67</td>
<td>3.3 d</td>
<td>88, 185, 300</td>
</tr>
<tr>
<td>Iodine-131</td>
<td>8.04 d</td>
<td>365</td>
</tr>
<tr>
<td>Iodine-123</td>
<td>13.2 h</td>
<td>159</td>
</tr>
<tr>
<td>Xenon-127</td>
<td>36.4 d</td>
<td>172, 203, 375</td>
</tr>
<tr>
<td>Xenon-133</td>
<td>5.25 d</td>
<td>81, 161</td>
</tr>
</tbody>
</table>

![Diagram of photon interactions inside a patient (ellipse) and collimation of the exiting photons before detection. Photon path a indicates a photoelectric interaction in which the photon is totally absorbed by an atom. Photon paths b and b' indicate a Compton scatter interaction in which the photon changes its direction of travel, with partial loss of its original energy. b, a scattered photon, and c, a primary, unscattered photon, pass through holes of the collimator and are detected by the radiation detector. b' and d are photons that are blocked by the collimator hole septa and are not detected by the scintillation camera.](image)

Figure 2. Diagram of photon interactions inside a patient (ellipse) and collimation of the exiting photons before detection. Photon path a indicates a photoelectric interaction in which the photon is totally absorbed by an atom. Photon paths b and b' indicate a Compton scatter interaction in which the photon changes its direction of travel, with partial loss of its original energy. b, a scattered photon, and c, a primary, unscattered photon, pass through holes of the collimator and are detected by the radiation detector. b' and d are photons that are blocked by the collimator hole septa and are not detected by the scintillation camera.

To multiplies tubes at its back. The detected photons are transformed into visible light (i.e., scintillations), which is converted into electrical signals by the photomultiplier tubes. The magnitudes of the electrical signals are proportional to the energies of the photons. A pulse height analyzer evaluates the energies of the detected photons and accepts only those that fall within a preset energy window, which is centered at the energy peak of the primary photons; thus, scattered photons and their adverse effects are rejected. The effectiveness of this scatter rejection depends on the energy resolution of the detector and the width of the energy window. Finally, a two-dimensional (2D) projection image of the three-dimensional (3D) radioactivity distribution is formed from the registered photon counts.

The half-lives and primary photon energies of the radionuclides commonly used in radiopharmaceuticals in scintigraphy and SPECT are shown in the Table. Among them, Tc-99m, with a primary photon energy of 140 keV, is the most popular.

### PHYSICAL AND INSTRUMENTATION FACTORS THAT AFFECT SPECT IMAGES

A number of factors substantially affect the quality and quantitative accuracy of SPECT images (5,6). They include physical factors, which are related to the interaction of photons emitted from the radiopharmaceuticals with matter inside the patient, and instrumentation factors, which are related to the SPECT imaging system. Figure 2 shows a schematic of photon interactions inside a patient and collimation of the exiting photons before detection. For photon energies commonly used in scintigraphy and SPECT, photoelectric and Compton scatter are the major interactions of radiation with matter, and no pair production interaction is possible due to the lower photon energies. In a photoelectric interaction, the photon is totally absorbed by an atom. In a Compton scatter interaction, the photon changes its direction of travel, with partial loss of its original energy.

Photon attenuation can be defined as the reduction of the number of primary photons that pass through a given thickness of material. Photon attenuation is caused by photoelectric and Compton scatter interactions in the material. The amount of attenuation depends on the incident photon energy and on the thickness and...
attenuation coefficient of the material. For example, the 140-keV photons from Tc-99m lose half of their original number in about 4.5 cm of water or soft tissue. As a result, photon attenuation substantially affects the quantitative accuracy of scintigraphic and SPECT images.

Photons that have been scattered before reaching the radiation detector provide misplaced spatial information about the origin of the radioactive source. Because the energies of the scattered photons are lower than those of the original or primary photons, scatter rejection is possible with use of an energy discrimination method. However, the energy resolution of a typical scintillation camera is approximately 10% at 140 keV. The finite energy resolution does not allow total rejection of the scattered photons. For example, the ratio of scattered to unscattered 140-keV photons detected by a typical scintillation detector is about 20%-50% in brain SPECT studies and about 30%-40% in cardiac and body SPECT studies.

The general shape of the scatter response function consists of a central peak surrounded by a broad tail. The tail portion of the scatter response function broadens as a function of the depth of the radioactive source, and its shape depends on the source location and the configuration of the patient's body. The shape of the function becomes increasingly asymmetric when the radioactive source is located closer to the edge of the body. If not compensated for, scattered photons contribute to inaccurate quantitative information and lower image contrast in SPECT reconstructed images without substantially degrading the spatial resolution.

In Figure 2, the photon paths b and c indicate a scattered photon and a primary photon, respectively, which pass through collimator holes properly and are detected by the scintillation camera. Photons may travel in directions that are intercepted by the collimator hole septa, indicated by the paths b' and d. The collimator is the major factor that determines the spatial resolution and detection efficiency of a conventional scintigraphic or SPECT system. The spatial extent of collimator-detector response (or collimator-detector response function) increases as a function of distance between the source and the detector and is usually symmetric. If not compensated for, collimator-detector response contributes to blurring, or loss of spatial resolution, and to asymmetry in the point response in the SPECT reconstructed images.

![Diagram of SPECT cameras](image)

**Figure 3.** Diagrams show typical configurations of commercial SPECT imaging systems. The increased number of scintillation cameras around the patient results in increased detection efficiency or improved spatial resolution.

### PLANAR VERSUS TOMOGRAPHIC TECHNIQUES

A planar scintiscan is a projection of a 3D radioactivity distribution onto a 2D plane. As a result, it is difficult to extract detailed information about an object of interest from the superposition of overlying or underlying radioactivity. The goal of image reconstruction methods in SPECT is to estimate accurately the 3D radioactivity distribution in vivo without the overlapping information. SPECT imaging systems are designed to acquire projection data (similar to those obtained with the conventional scintigraphic method) accurately and efficiently from multiple views around the patient. Image reconstruction methods are applied to the projection data to generate SPECT images of the 3D distribution of radioactivity.

### SPECT INSTRUMENTATION

SPECT imaging systems can be divided into those found in the commercial market and those in research laboratories (3,7). The trend is to place larger or greater numbers of detectors around the patient to increase detection efficiency. Figure 3 shows typical configurations of commercially available SPECT systems. The early SPECT systems are based on a single scintillation camera mounted on a rotating gantry. Recently, SPECT systems with two, three, and four cameras have become available in the market. The increased number of scintillation cameras around the patient results in increased de-
tection efficiency. The increased detection efficiency can be traded for improved spatial resolution, resulting in improved SPECT image quality.

Collimation is the most important component that determines both spatial resolution (blurring) and detection efficiency of scintigraphy and SPECT imaging systems. The width of the collimator response function (degree of blurring) depends on the particular collimator design. In general, a higher-resolution (less blurring) collimator results in lower detection efficiency (higher noise level) in the detected image and vice versa.

Similar to the collimators used in planar scintigraphy, parallel-hole collimators are most commonly used in SPECT. Converging-hole collimators (eg, fan beam, cone beam, and astigmatic collimators) have also been developed for SPECT (8). Fan-beam and cone-beam collimators provide approximately 1.5 and 2 times higher detection efficiency, respectively, than a parallel-hole collimator with the same spatial resolution. However, the increased detection efficiency is accompanied by a decrease in the field of view. As a result, converging-hole collimators are limited to the imaging of small organs (eg, the heart) or body parts (eg, the head). Also, the use of these collimators in SPECT requires special data acquisition strategies and image reconstruction algorithms.

CONVENTIONAL SPECT RECONSTRUCTION METHODS

The goal of SPECT image reconstruction methods is to estimate the true radioactivity distribution in vivo from the measured projection data. If the measured projections are simple sums of radioactivity along the projection rays (ie, if the effects of attenuation, scatter, and collimator-detector response can be ignored), conventional image reconstruction methods such as the popular filtered backprojection algorithm will generate quantitatively accurate SPECT images (9). Also, the filtered backprojection algorithm allows selection of smoothing filters to suppress image noise and deconvolution filters to provide partial compensation of blurring caused by the collimator-detector response (10).

However, in practice, the effects of attenuation, scatter, and collimator-detector response in SPECT are substantial, and the assumption that the measured projection data are simple sums of radioactivity along the projection rays is severely violated. When the filtered backprojection algorithm is used to reconstruct the measured SPECT data, the quality and quantitative accuracy of the reconstructed images will be severely degraded.

To demonstrate the effects of the factors that degrade SPECT images, a simulated data set was generated from CT scan data of a patient. Figure 4 shows a sample section from a set of multiple CT images obtained through the chest region of a patient; the CT number distribution from the corresponding section is shown in Figure 4a. The CT number distribution can be transformed to an attenuation coefficient distribution corresponding to the primary photon energy of Tl-201. The information is then used to simulate a 3D phantom with uptake distribution of Tl-201 in various organs, and the corresponding section is shown in Figure 4b.

From the phantom shown in Figure 4, projection data from 180° around the patient (45° left posterior oblique to 45° right anterior oblique) were simulated, both without any degrading effects (ideal imaging system) and with
the effects of attenuation, scatter, collimator-detector response, and statistical noise fluctuations. The images in Figure 5 were obtained by using the conventional filtered backprojection reconstruction algorithm to reconstruct the different sets of simulated projection data. They demonstrate the various degrading effects on the SPECT images.

## QUANTITATIVE SPECT RECONSTRUCTION METHODS

To obtain good-quality and quantitatively accurate SPECT images, the image-degrading factors described earlier must be compensated for. Quantitative SPECT reconstruction methods consist of two main components: (a) an algorithm for image reconstruction from multiple projections and (b) methods to compensate for the image-degrading factors (5).

The characteristics of the degrading factors that affect SPECT are complex. Examples are the nonuniform attenuation distribution in the chest region, the spatially variant (or nonstationary) collimator-detector response, and the asymmetric scatter response whose magnitude and shape depend on source position. These characteristics make it difficult to compensate for the image-degrading factors exactly. Compensation methods have been developed that take advantage of the efficient filtered backprojection algorithm. However, use of these methods is usually based on certain assumptions about the degrading factors, and they can achieve only partial compensation.

Quantitative SPECT reconstruction methods accurately take into account the image-degrading effects (5). A class of image reconstruction methods is based on iterative algorithms, which allow incorporation of models of the imaging process and which provide accurate compensation for the image-degrading factors. An example is the iterative expectation maximization (EM) algorithm that estimates the maximum likelihood solution of the reconstruction problem for measured projection data whose noise fluctuations are distributed according to the Poisson statistics.

A typical iterative reconstruction algorithm starts with an initial estimate of the object source distribution. A set of projection data is
generated from the initial estimate by using a projector that models the imaging process and the image-degrading factors. The generated projection data are compared with the measured projection data at the same projection views, and their differences are calculated. On the basis of specific statistical criteria, the differences are used to update the initial image estimate. The updated image estimate is then used to recalculate a new set of projection data that are again compared with the measured projection data. The procedure is repeated until the differences between the calculated and measured projection data are smaller than a preselected small value. Compensation for the image-degrading effects is achieved by modeling the imaging processing in the projection and back-projection steps of an iterative reconstruction algorithm.

Attenuation is the most important factor affecting the quality and quantitative accuracy of SPECT images. Attenuation compensation methods can be divided into two groups. The first group of methods assumes that the attenuation inside the patient is uniform. These methods can be used for SPECT studies of the brain and abdominal regions for which the assumption of uniform attenuation is fairly accurate or a good approximation. The second group deals with cases in which the attenuation coefficient distribution inside the patient is nonuniform. This second group of methods is used for cardiac and lung SPECT studies because the chest region consists of tissues, such as lung, soft tissue, and bone, that have different attenuation coefficients.

Methods of compensating for uniform attenuation in the first group can be further divided into two categories. The first category of methods is based on analytic solutions of the reconstruction problem. They provide exact compensation for the uniform attenuation. However, these methods tend to amplify noise in the reconstructed images. The other category of methods is derived by making simple assumptions about the uniform attenuation problem. The most popular method is the Chang algorithm in which the average attenuation factors for each reconstructed image pixel are calculated based on knowledge of the body contour and the average attenuation coefficient. The attenuation-compensated image is obtained by multiplying each pixel of the image reconstructed with filtered backprojection but without attenuation compensation by the average attenuation factor of the same pixel.

Because an analytic solution to the nonuniform attenuation problem has not been found, the attenuation compensation methods in the second group consist of two steps. The attenuation coefficient distribution (or map) through the body region to be imaged is first determined separately. The attenuation coefficient map is then incorporated in an iterative reconstruction algorithm for accurate attenuation compensation.

Recently, compensation for nonuniform attenuation distribution in cardiac SPECT studies has been an active area of research. The attenuation distribution inside the patient's body is obtained by transmission CT methods by using the SPECT imaging system. An external radionuclide source is used to provide radiation that is transmitted through the patient and the detector in the same SPECT system is used for detection. The attenuation-compensated cardiac SPECT images have been shown to provide much improved quantitative accuracy and image quality in terms of greatly reduced image artifacts and distortions, although the spatial resolution of the image has not been improved. Furthermore, attenuation compensation can reduce the artifactual myocardial count density decrease due to diaphragmatic and breast attenuation effects. This has the potential of reducing the number of false-positive and false-negative diagnoses.

Methods to compensate for scatter can be grouped into three general categories. In the first category of methods, it is assumed that the scatter component contains no useful information. The scatter component is estimated and subtracted from the measured data or from the reconstructed images. Such methods include those that use dual- and multiple-energy windows and those that fit curves to the energy spectral data. The second group of methods utilizes a weighted sum of data obtained from multiple-energy windows. The weighted sums are determined by a scheme to maximize the signal-to-noise ratio of the weighted image. The third group of methods recognizes that the scatter component does contain useful image information, even though it is blurred by the scatter response function. In an efficient but approximate compensation method that implements this third approach, the scatter response function is estimated and the blurred image information is recovered by applying deconvolution techniques. An accurate implementation of the third approach models the exact scatter response function in the reconstruction method (e.g., those using iterative algorithms) at a cost of long processing time.
Methods to compensate for collimator-detector response can be grouped into two general categories. The first category of methods assumes an average collimator-detector response and incorporates it in a deconvolution filter, for example, the Metz or the Wiener filter. The methods provide approximate compensation for the spatially variant collimator-detector response. A more accurate compensation approach models the spatially variant collimator-detector response function and incorporates it in the reconstruction methods, which can be analytic or iterative.

Quantitative SPECT reconstruction methods that compensate for the image-degrading factors can be implemented in 2D or 3D (11). In the 2D implementation, a 2D model of the imaging process is used in section-by-section 2D image reconstruction and compensation. The final 3D image is formed by stacking the multiple 2D reconstructed images. In the 3D implementation, a 3D model of the imaging process is used in direct 3D image reconstruction and compensation to form the final 3D reconstructed image. Although the 3D implementation methods provide the most accurate compensation for image degradation, they require much longer processing time compared with that needed for the 2D implementation methods.

**EXAMPLES OF QUANTITATIVE SPECT IMAGES**

To demonstrate the effectiveness of the quantitative SPECT reconstruction methods, SPECT was performed on a 3D Hoffman brain phantom that simulates the radioactivity uptake in a normal blood flow study of the brain. The attenuation coefficient in the phantom is uniform throughout. A 2D image section of the phantom was reconstructed with use of the filtered backprojection algorithm without any compensation; the resulting image was of poor quality and had lower count density toward its center due to the attenuation effect (Fig 6).

Figure 7 shows results from different 2D and 3D implementations of the quantitative SPECT reconstruction methods on the same 3D data used in generating the image shown in Figure 6b. Images were obtained with (a) 2D approximate attenuation compensation by using the Chang method and approximate collimator-detector response compensation by using a 2D Metz filter, (b) 2D accurate attenuation and collimator-detector compensation by using the iterative EM algorithm with exact modeling of the degrading effects, (c) 3D approximate attenuation compensation by using the Chang method and approximate collimator-detector
A major clinical application of quantitative reconstruction methods has been to compensate for attenuation in cardiac SPECT studies. Figure 8 shows sample transaxial transmission CT images through the chest of a patient. The images were obtained by using a pair of gadolinium-153 scanning line sources in combination with an Optima SPECT system (GE Medical Systems, Milwaukee, Wis). The images were used as attenuation coefficient maps for compensation of attenuation. Figure 9 shows TI-201 SPECT images reconstructed from the corresponding transaxial sections in Figure 8 by using the conventional filtered backprojection reconstruction algorithm without attenuation compensation.
SPECT images reconstructed from the same transaxial CT scans shown in Figure 8. The conventional filtered backprojection reconstruction algorithm without attenuation compensation was used. The reconstructed images were postprocessed by using a Butterworth filter with order 8 and a cutoff frequency of 0.39 cycles per centimeter. Image artifacts and distortions from the attenuation effects are evident.

Figure 10 shows reconstructed images of the same section shown in Figure 9, except that 50 iterations of the EM algorithm with attenuation compensation and the attenuation maps shown in Figure 8 were used in the reconstruction. The attenuation-compensated images were postprocessed by using the same Butterworth filter employed in the filtered backprojection reconstructions. The images in Figure 10 show much improved image quality in terms of reduced image artifacts and distortions. The images shown in Figure 11 are similar to those in Figure 10, except that projection data were compensated for scatter by using the convolution subtraction method before image reconstruction. The reconstructed images show improved image contrast relative to those in Figure 9.

**CONCLUSION**

In conclusion, SPECT combines conventional scintigraphic techniques and image reconstruction techniques. SPECT images provide detailed 3D information of the distribution of radiopharmaceutical uptake that reflects functional information about the patient. However, SPECT images are degraded by noise, attenuation, scatter, and collimator-detector blur. Quantitative SPECT reconstruction methods consist of image reconstruction algorithms and techniques to compensate for image-degrading effects. These reconstruction methods provide both improved quality and quantitative accuracy in the final image. The future of SPECT depends on advances in radiopharmaceuticals, instrumentation, quantitative reconstruction methods, and successful clinical applications.
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