Attenuation correction in cardiac positron emission tomography and single-photon emission computed tomography

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Quantitation in cardiac positron emission tomography (PET) and single-photon emission computed tomography (SPECT) depends on being able to correct for several physical factors that tend to distort the data. One of the most important of these corrections is the correction for attenuation. For PET, cardiac attenuation correction is a reality, although certain problems remain to be solved. For SPECT, recent developments in gamma camera hardware and reconstruction methods have finally made it possible to attempt attenuation correction in a clinical setting. This article reviews the methods available to perform attenuation correction in both PET and SPECT, with emphasis on the commonality between the problems encountered and solutions proposed for each modality. (J NUCL CARDIOL 1995;2:246-55.)

Key Words: Cardiac position emission tomography · single-photon emission computed tomography · attenuation correction

It is widely recognized that the ability to correct cardiac emission data for the effects of attenuation may dramatically improve the clinical utility of the resulting reconstructed images.[1] In positron emission tomography (PET), such attenuation corrections are possible, although certain problems remain to be solved. In single-photon emission computed tomography (SPECT), the situation is more difficult. Many schemes have been proposed to measure attenuation values, and at least one of these methods is now available commercially. Much work remains before attenuation correction becomes as practical with SPECT as it is with PET, and validation studies have yet to be done. Nonetheless, hardware to measure SPECT attenuation maps and software to use those maps to perform attenuation correction are now on the market. This then seems an opportune moment to review the basic principles of attenuation correction and to summarize the approaches that have been proposed for performing attenuation correction.

This review will concern itself with the approaches taken to address the three basic problems of attenuation correction: (1) how to determine the attenuation correction factors, (2) how to use these correction factors to reconstruct the tomographic slice properly, and (3) how to address the practical problems (e.g., patient motion and noise) that occur when correcting tomographic images for attenuation. It should be remembered that quantitation in SPECT can be achieved only if attenuation correction is combined with other required corrections, most notably scatter correction. This article will focus only on the methods used to perform attenuation correction in both PET and SPECT. Even for the reader whose sole interest is in SPECT scanning, it is useful to understand how the process works for PET. Several of the methods used in SPECT are similar to (and may have been inspired by) those employed in PET. SPECT attenuation correction is a far more difficult problem than PET attenuation correction. The solutions proposed for SPECT are much easier to understand if one first learns how the problem has been approached in PET.

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THE ATTENUATION PROCESS

Attenuation produces dramatic effects in both PET and SPECT cardiac imaging. The 511 keV photons used in PET are attenuated less per centimeter of soft soft tissue than are the photons typically used in SPECT. The half-value thickness—the number of centimeters of tissue needed to stop half the photons—is 7.2 cm for 511 keV photons and 4.6 and

3.8 cm for ^{99m}Tc and ^{201}Tl, respectively. In SPECT imaging, however, the photons need travel only from the heart to the camera, whereas for PET the pair of coincident photons emitted from within the heart must traverse the entire body thickness together to reach the opposing pair of detectors in the imaging device. Therefore the effects of attenuation are large for both modalities. Typically only about 1 of 10 to 1 of 40 photons (or pairs of photons) is able to reach the imaging device and produce a usable count, depending on the position of the detector and body habitus. Figure 1 shows a sketch made from a transaxial magnetic resonance imaging slice at a midventricular level, along with some typical distances (in millimeters) that a photon might have to travel to reach a gamma camera in a left anterior oblique position (or to reach the corresponding pair of detectors in a PET camera). For a single-photon emitter (e.g., ^{99m}Tc), a decay from location A in the myocardium (Figure 1) might have to traverse 44 mm of soft tissue to reach the gamma camera (at this angle), whereas a decay from location B would have to travel much farther. For PET the total path length is considerably longer (128 mm in the lung and 164 mm in soft tissue in this patient at this angle), because both coincident photons must travel across the entire body to reach their respective detectors. The total attenuation is actually greater for PET than for SPECT in this particular direction. However, it is important to note that the attenuation in the PET case is independent of whether the photon was emitted from location A or B in the myocardium. The total attenuation is the same for either because the total distance traversed by the pair of photons is the same no matter where along the path of the photons (called the projection line) the decay took place. This is why it is comparatively easy to perform attenuation correction in PET. In PET the photons emitted along any particular projection line may be corrected for attenuation without knowing how deep in the body the decay took place. Only the total attenuation through the whole body along that direction must be known. In SPECT, on the other hand, one needs to know exactly how deep along a projection line the radioactive decay took place to correct that projection line for attenuation. This is the "catch 22" that has plagued SPECT since its inception. You cannot create an accurate tomographic image of the distribution of radioactivity in the body unless you perform attenuation correction; however, you cannot correct the projection data for attenuation unless you know exactly how the radioactivity is distributed. Fortunately, there are now several computational approaches for getting around this catch 22. These, combined with the other methods discussed below, have put us at the threshold of being able to perform accurate SPECT imaging.

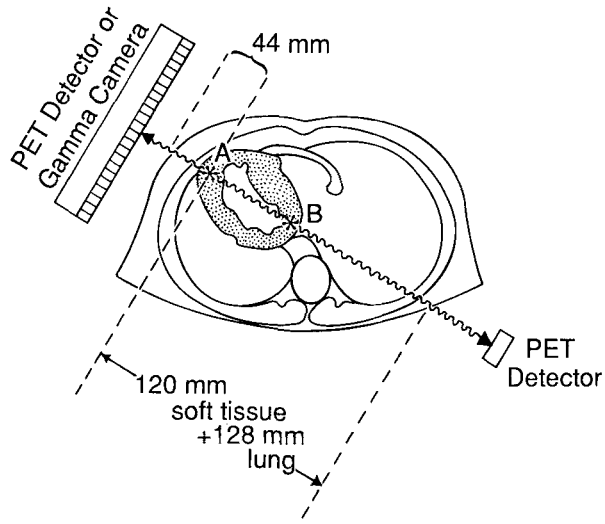

Figure 1. Thickness of tissue through which SPECT or PET photons must traverse at one particular projection angle. Two possible emissions are shown, labeled *A* and *B*.

MEASURING THE ATTENUATION PROPERTIES OF THE BODY

Positron Emission Tomography. In PET it is in principle easy to measure the attenuation correction factors needed to correct the counts seen by each pair of detectors. All one need do is measure the total attenuation occurring along each possible projection line through the body. Because in PET the attenuation does not depend on where along the projection line the source is located, it can just as well be located outside the body, as shown in Figure 2. The total coincident counts measured with the source, divided by the total counts along the same projection line when the patient is removed (called the "blank" scan), gives the fractional attenuation along that projection line through the body. The inverse of this factor is the correction factor needed. All possible projection lines must be corrected in this way. The measurement must be repeated as the source is moved completely around the gantry with the patient and bed present (the transmission scan) and with no patient or bed (the blank scan). The blank scan divided by the transmission scan yields the array of attenuation correction factors needed to correct each projection line. Once each projection line has been corrected for attenuation (after correcting for scatter), the emission data may be reconstructed into an attenuation-corrected emission image. It is also possible to reconstruct the array of attenuation correc

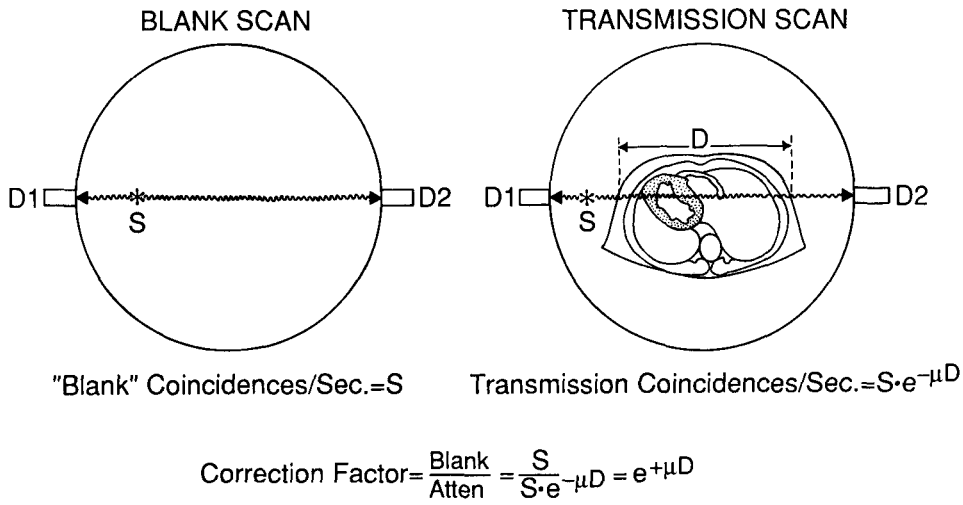

Figure 2. Illustration of attenuation correction in PET. Counts in blank scan divided by those in transmission scan give attenuation correction factor. Source is shown in one particular position, but to produce correction factors at all angles it must be rotated around body.

tion factors to produce an image in which the intensity of each pixel is proportional to the attenuation value (μ) at that point—the equivalent of a computed tomographic scan at 511 keV. However this latter step is not necessary for the reconstruction of the PET emission data.

Because nearly all PET scanners are multislice scanners, a rod source rather than a point source is used. ^{68}Ge (288-day half-life) is the radioisotope usually used. The rod is oriented along the long axis of the gantry and, to be outside the patient, is located near the detectors. This can result in severe dead-time problems if the source is too hot. On the other hand, because the patient blocks so many of the photons, one would like the source to be as hot as possible to get sufficient statistics along each projection line. It has been shown that there is an optimum activity for the rod source.[2] For a given transmission scan time, activities below this optimum amount produce noisier data, whereas increasing the activity above this value increases radiation exposure to the patient without a corresponding increase in useful counts, because of the losses caused by dead time.

Because the count rate is so high for the detectors nearest the rod source, dead time may be a severe problem, especially for septa-less PET scanners (so-called three-dimensional PET scanners) or large crystal PET scanners.[3] One proposed solution to this problem[4] is to abandon the use of coincident events for attenuation measurements and instead record only the counts in the detectors farthest from the line source. The colinearity between the detected single event and the rod source defines the projection line. Scatter and the differences in blank and transmission dead times must be accounted for, but this approach may yield higher count rates with weaker rod sources than with coincidence counting.

Scatter can artificially decrease the measured attenuation by increasing the number of photons that appear to make it through the patient (Figure 3). Random events produce a similar effect. Both effects are greatly reduced by electronically excluding all those events occurring in detectors that are not colinear with the rod source.[2,5,6] A variant of this electronic collimation is also used in SPECT, as will be seen below. It must be remembered that because the number of photons making it through the body is so low, even a small number of erroneous scattered or random events in the transmission scan can change the measured attenuation factors significantly.

Single-Photon Emission Computed Tomography (SPECT). The techniques for measuring attenuation maps in SPECT are similar to those for PET. In SPECT each projection is just a planar gamma camera view. Originally, planar transmission scans were obtained with an uncollimated flood source (either a liquid fillable source or a solid ^{57}Co source). However, the amount of scatter was so large that the μ values obtained (from the log of the ratio of the blank to the transmission scan) were systematically low: typically 0.12 cm^{-1} for soft tissue at 140 keV instead of the true value of approximately 0.15 cm^{-1}. To correct this, the source must be collimated.[7] This drastically reduces the radiation exposure to the patient and eliminates nearly all scatter, yielding μ values nearly identical to the true values. The

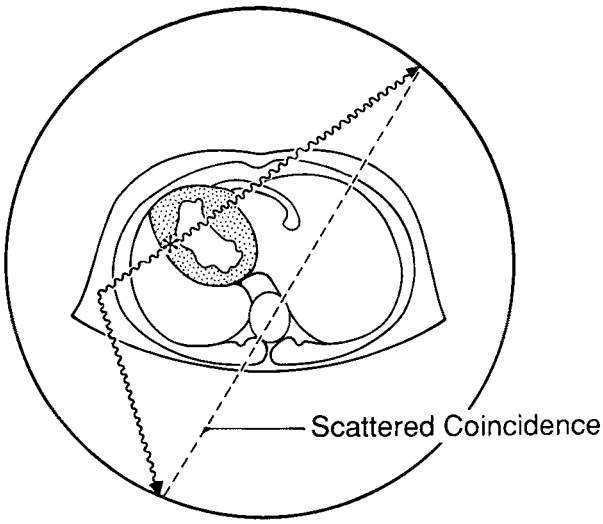

Figure 3. Illustration of how scattered event can produce coincidence line far from where true decay occurred. (From Bergmann SR, Sobel BE, eds. Positron tomography of the heart. Mount Kisco, New York: Futura Publishing, 1992.)

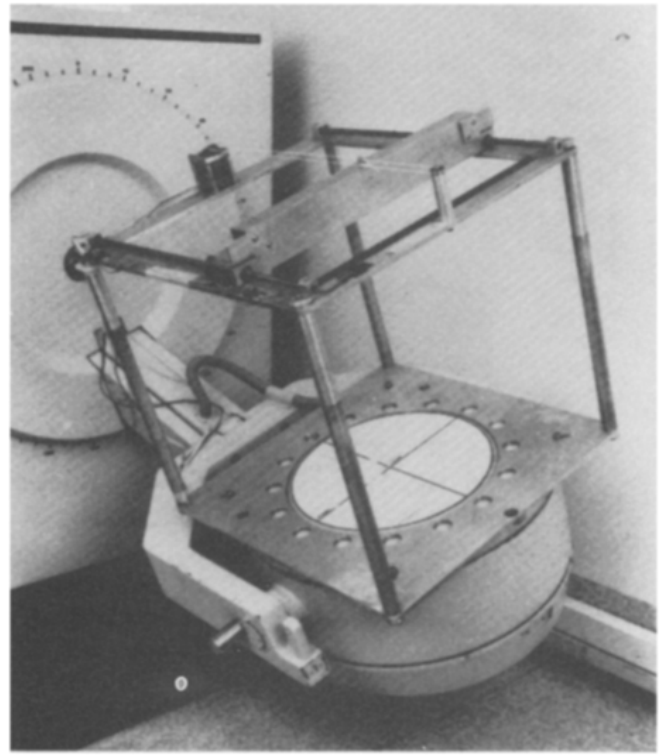

Figure 4. One of first scanning line source devices for transmission correction in SPECT. (From Tan P, Bailey DL, Meikle SR, Eberl S, Fulton RR, Hutton BF. J Nucl Med 1993;34:1752-60.)

collimator can be a higher sensitivity, lower resolution collimator than that used on the gamma camera. Some care must be taken to align the two collimators.

There are several practical difficulties in using a collimated flood source. First, they are difficult to handle (and fill, in the case of liquid sources), because very large activities are necessary (hundreds of millicuries) to get reasonably short attenuation scan times (note that even at this level of activity the exposure to the patient is very small, because of the collimator). Second, the collimator's flood source assembly can be quite heavy, and mounting it to a gantry is difficult. Third, it is not practical to use such a system for three-headed systems (although in principle it could be used for two-headed, 90-degree systems). Finally, unless very large activities are used in the source, it would be impossible to use such a source in the presence of emission contamination.

One practical alternative to the collimated flood source is to use a collimated line source, which scans across the camera field of view. In one such arrangement, the line source is narrowly collimated (Figure 4) to reduce scatter and patient exposure, and the line source assembly is driven by a motor that translates it across the field of view.[8] The line source can be made of a relatively long-lived isotope (e.g., ^{153}Gd, ~99 keV, 240-day half-life, or ^{57}Co, 120 keV, 271 days) and shielded for easy handling. The duration the source spends at each position must be either uniform or monitored accurately, so that the blank scan and the transmission scan can be made comparable before division. Again, the source activity must be high to get sufficient transmitted counts through the human thorax in as short a time as possible. One may think of the collimated line source as the same as one narrow line of the parallel-hole collimated flood source but without all the attendant weight (with the added difference that the collimation is only in one direction in the case of the line source). If the rod source activity were the same as that in the entire flood source, it would produce a transmission scan of the same statistical certainty in roughly the same time as would the collimated flood source.

During the blank scan, or while the line source is outside the boundaries of the body, the count rate to the camera may be high for a high activity source, which may make the camera count rate performance a limiting factor.

The scanning line source is not practical for three-headed cameras. Therefore another approach that uses a line source has been proposed (and in fact is available commercially).[9-11] In this approach a line source is located at one of the gaps between the three camera heads (Figure 5). The opposing camera head is then used to detect the transmitted photons. If a parallel-hole collimator were used on this head, it

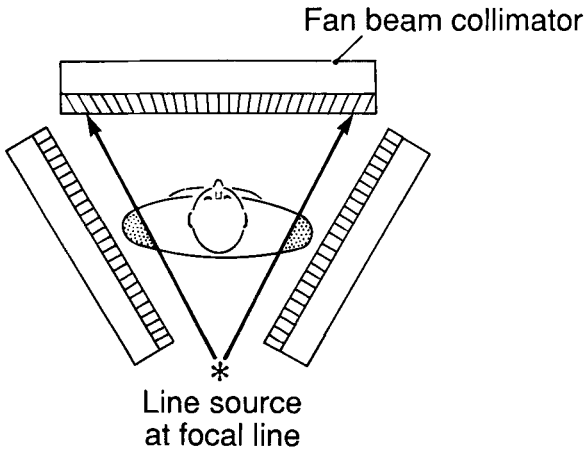

Figure 5. Fan beam collimator and coarsely collimated line source, used for attenuation correction with three-headed gamma cameras. Note *shaded* portion of patient's body will be truncated (i.e., not visualized) by fan beam collimator. (Similar to a figure by Tung CH, Gullberg GT, Zeng GL, Christian PE, Datz FL, Morgan HT. IEEE Trans Nucl Sci 1992;39:1134-43.)

would be necessary to scan the line source. Instead, a fan beam collimator is used. This fan beam collimator is designed such that all the lines of sight through the collimator holes converge on the line source. Therefore a single stationary line source can be used to irradiate the entire camera head through the fan beam collimator. Because every set of collimator holes sees the line source at once, only a relatively small amount of activity is needed to yield a fairly high count rate. It is obviously important that the source remains at the focal point (actually the focal line) of the fan beam collimator. If a body contour orbit is used during the SPECT acquisition, however, the heads may move in and out of the center of rotation. If so, some sort of mechanical controller is needed to adjust the position of the source so that it is kept at the focal line during the orbit. The source is shielded so that it exposes only the field of view of the transmission camera head and results in a very low radiation exposure to the patient.

One problem with the fan beam method is a loss of data called "truncation." Because the fan beam collimator focuses down to a line in about 50 cm, the camera field of view diminishes as a function of distance from the collimator (Figure 5). It is possible that a portion of the patient's body will not intersect the fan beam and so will be out of the camera's field of view at certain projection angles (e.g., at the anterior position). This can lead to missing data and result in artifacts, called truncation artifacts, when reconstructing the attenuation image. For SPECT, as will be seen below, the attenuation image is necessary for most of the schemes that correct the emission data for attenuation (as opposed to PET in which only the attenuation correction factors are necessary). Several schemes have been proposed to account mathematically (in an approximate way) for this truncation.[12-14]

One final method that has been proposed is to use not a fan beam collimator focusing on a transmission line source but rather a cone beam collimator focusing on a transmission point source.[15-17] However, the truncation problems for this geometry are even more severe than for fan beam collimation, and as yet this method has not proved to be clinically practical.

Scaling the Energy of the Transmission Source in SPECT. When the transmission scan is performed with a radioisotope different from that used for the emission scan, the attenuation coefficients must be corrected to the value appropriate for the emission gamma ray energy. It is usually assumed that the ratio of μ values for two different energy photons does not depend on the attenuating medium. For the photon energies typically used in SPECT and PET, this assumption is reasonable for soft tissue and lungs. However, for bone the approximation is expected to be less accurate because of the high atomic number of calcium.[11] In clinical practice, for energies above 100 keV, the errors caused by making the approximation are probably acceptably small.

An alternative approach is to measure attenuation coefficients experimentally for different materials at both energies of interest and thereby deduce an empirical relationship between the μ values.

It has also been suggested that one could use a CT image to compute the map of attenuation coefficients. In this case a relationship converting Hounsfield's units to attenuation coefficients at the emission energy must be derived. When only one CT image obtained with an electron beam of a particular energy is used, the conversion is not uniquely defined because the attenuation coefficients depend on both density and atomic number. Some assumptions are therefore necessary to get a conversion formula.[18,19] The effect of errors in the conversion of Hounsfield's units to attenuation coefficients on the accuracy of the attenuation-corrected images has been reported.[19-21]

Postinjection Transmission Scanning. It is frequently necessary, in both PET and SPECT, to make an attenuation measurement in a subject after the radiopharmaceutical has already been injected. This is often necessary when imaging pharmaceuticals with long uptake periods, when performing repeat scans on patients, or when consecutively imaging adjacent regions of the body. In PET the energy of the transmission source is identical to the energy of the emission, making separation of the two by energy

discrimination impossible. However, the electronic requirement of colinearity with the rod, described above, greatly reduces the fraction of emission counts seen concurrently when performing transmission scanning (3% to 10%, depending on how tightly the colinearity requirement is enforced by the electronics). Still, because of the large attenuation of the thorax during a transmission scan, even a small number of emission photons can greatly distort the computed attenuation correction factors. Therefore to perform an accurate postinjection PET transmission scan, one must also have an estimate of the emission data present (e.g., from a previous or subsequent scan) and use that estimate to correct the emission-contaminated transmission data. As long as the fraction of emission counts contaminating the transmission scan is small, some error can be tolerated in this correction. The accuracy of performing PET postinjection transmission scanning has been well validated for brain imaging.[22,23] Considerably less data are available as to the efficacy of the method for thoracic imaging, but it should be possible in principle there as well.

Postinjection transmission correction is also possible with SPECT but has been studied less than for PET.[10] With the scanning line source method, contamination from the emission counts can be reduced with electronic collimation. This collimation turns off all parts of the face of the gamma camera that are not being exposed by the transmission source. In SPECT it is possible to use a transmission source whose energy differs from that of the emission activity. Energy discrimination is therefore possible to reduce further emission contamination of the transmission data. This contamination would be best eliminated by use of a transmission radioisotope whose energy was higher than that of the emission data (e.g., ^{57}Co or ^{153}Ge when scanning ^{201}Tl). In this way, downscatter of the emission counts will not contaminate the transmission data. If the injected activity is ^{99m}Tc (or a higher energy radioisotope), the problem is more difficult, but other transmission radioisotopes (e.g., 166 keV ^{139}Ce or 159 keV ^{123m}Te) are being investigated.[24]

Simultaneous Transmission/Emission Scanning. Ideally, for either PET or SPECT imaging, one would like to acquire the transmission scan simultaneously with the emission data. If this were possible, although it might not significantly reduce the total scanning time, it would eliminate the potential effects of patient motion occurring between the time of the transmission scan and the emission scan.

In PET, simultaneous transmission/emission scanning is possible but as yet has not proved practical. In SPECT, there is evidence that simultaneous transmission/emission scanning may be both practical and accurate, although possibly with some sacrifice of either scan time or noise.[8,9,25,26] The simplest approach with a three-headed or two-headed 90-degree camera system would be to dedicate one of the heads to the acquisition of transmission data and use the other(s) for the acquisition of emission data. Obviously this increases the emission scan time by a factor related to the number of heads lost to emission scanning. In this mode it may be possible to use the same isotope for transmission as for emission. Because the photons must scatter through approximately 90 or 120 degrees (for the two-headed or three-headed configurations, respectively), the contamination of the emission data by the scattered transmission photons, or vice versa, can be reduced by energy discrimination. Alternatively, all heads can be used for acquisition of emission data while simultaneously one head acquires transmission data as well. In this case, isotopes of different energies must be used for emission and transmission and corrections made for scatter of transmission photons into emission data and vice versa. The use of several combinations of emission and transmission radioisotopes has been reported,[8,11,27] and various procedures for estimating and subtracting the contaminant image have been described.[25,26] It is important to note that the subtraction of contamination itself adds noise to the images, and the character of the resultant noise may no longer be Poisson. Consequently, scan time may need to be prolonged, and reconstruction algorithms that assume a Poisson noise distribution must be used with care.

USE OF THE ATTENUATION MAP TO CORRECT THE EMISSION DATA

In PET the acquisition of a reliable transmission map makes an accurate attenuation correction possible (ignoring practical issues such as motion and noise). For each projection line, the attenuation correction factor is given by the ratio of the blank scan to the transmission scan. After the emission scan, each projection line is multiplied by its attenuation correction factor, resulting in the same projections as would have been obtained had there been no attenuation. The corrected projection data can then be reconstructed to produce an attenuation-corrected emission image.

In SPECT the situation is much more complicated because one must know both the attenuation map and the emission distribution to apply an accurate attenuation correction. In other words, even

if one were given the planar projections and the attenuation map, no exact method for attenuation compensation has been found. As a result, many approximate correction methods have been proposed.

If one assumes that attenuation is uniform in the part of the body being scanned (i.e., that the attenuation coefficient is the same everywhere), the situation is much simpler. Indeed, in that particular case, theoretical solutions for attenuation correction exist.[28-30] However, for the thorax, unlike other parts of the body, the assumption of uniform attenuation is incorrect and its use can lead to severe artifacts in the corrected images.[31,32]

Other approaches, some similar to those used for PET, have been proposed. These involve either modifying the projections before the reconstruction to account for attenuation (as is done in PET) or modifying the reconstructed slices after reconstruction. The former are called prereconstruction corrections and the latter postreconstruction corrections. Several different prereconstruction corrections have been proposed. Some assume uniform attenuation,[33-35] whereas others deduce the correction factors from the map of the attenuation coefficients.[19,36] For instance, it has been proposed to estimate the attenuation factor affecting a projection line by computing the average of all attenuation factors along that projection line.[19] This is of course only approximately correct, because the accurate correction factors would require one to know both the attenuation coefficients and the activity distribution along that line. Similar ideas have been suggested for postreconstruction correction methods. For example, each pixel of the reconstructed volume (or slice) could be divided by a correction factor, such as the average overall projection lines of the attenuation factors affecting that pixel.[37] Although such corrections are easy to implement, because of the relatively crude approximations they entail they cannot in general compensate properly for nonuniform attenuation, even if an accurate map of attenuation values is used.[7,38]

None of the methods described above addresses the underlying cause of the SPECT attenuation correction problem (i.e., the catch 22 nature of the problem) described above. A more natural solution would be to use an iterative procedure. The idea is to first make a rough estimate of the activity distribution and then progressively improve this estimate by comparing, at each iteration, the projections one would have measured if the estimate were accurate with the projections actually measured. A first estimate of the activity distribution is obtained, with either no attenuation correction or a prereconstruction or postreconstruction correction. This first estimate is then reprojected through the measured attenuation map (i.e., one computes the projections that would have been obtained had this first estimate of the activity distribution been correct, given the measured attenuation map). Of course, these computed projections are not exactly the same as the projections actually acquired because the estimated activity distribution is only an approximation of the true activity distribution. However, the difference between these calculated and measured projections (called "error projections") gives an insight into what is wrong with the initial estimate of the activity distribution. With the error projections, one can make a new, revised estimate of the activity distribution, which corrects for the observed errors. This newly revised estimate of the activity distribution can then be reprojected again and the entire process repeated. Several methods that use this principle have been described.[19,20,37,39-42] These iterative correction methods can be combined with any reconstruction algorithm, and most of them use the standard method of filtered back-projection. The iterative methods usually out perform the prereconstruction or postreconstruction corrections, and they require only a few iterations (usually less than five) before they give results close to the true distribution. Consequently, they are relatively fast. Their main shortcoming is that it has not been proved mathematically that they will necessarily converge to the correct answer, even with an infinite number of iterations.

The idea of an iterative estimate of the activity distribution is appealing because it appears to address directly the catch 22 issue. There are, however, very different iterative schemes in which the basic reconstruction process itself is performed iteratively (rather than by filtered back-projection). The principle of iterative reconstruction involves mathematically modeling the imaging process (i.e., how the projections relate to the activity distribution). This model is then used to solve for the activity distribution given the observed projections, in an iterative way. When the attenuation map is known, attenuation can be modeled as part of the imaging process, and the well-known iterative reconstruction methods (e.g., ML-EM and algebraic reconstruction techniques[43,44]) can be used.[32,45-49] In fact, this is the most general approach to the reconstruction problem because, unlike the filtered back-projection, it can allow for not only attenuation but also scatter and detector response. In that respect it seems to be the ideal approach. However, as they are commonly applied, these approaches are usually more computationally demanding than filtered back-projection, and many

problems remain such as the optimization of the involved parameters and the choice of the stopping criterion for the iterative procedure. Still, because these methods offer a more general approach to the whole reconstruction problem, they may one day become the standard for quantitative SPECT. At the present time, however, the iterative filtered back-projection methods appear to be the most clinically practical.

PRACTICAL PROBLEMS IN COMPUTING ATTENUATION-CORRECTED IMAGES

Patient Motion. In both PET and SPECT, when separate transmission and emission scans are performed, it is possible that the patient may move slightly between the transmission scan and the emission scan. In that case the correction factors will be applied to the incorrect emission data. For PET, some of the effects that can be produced by such inadvertent motion have been studied by McCord et al.[50] They showed that 5 mm of lateral motion, or 1 cm of z-axis motion, or 5 degrees of rotational motion could change the apparent regional distribution of activity around the myocardium by approximately 10%. The biggest effects occur at lung/myocardial interfaces, where attenuation changes rapidly. The septal wall, on the other hand, is surrounded by regions with similar attenuation properties (the left ventricular cavity and the right ventricle) and so is affected less by such translations. Patient motion can therefore cause the free wall/septal ratio to be incorrect.

Similar effects have been predicted to occur if misalignment between the transmission and emission scans occurs in SPECT,[51] but the problem has not been studied as thoroughly. One approach that has been proposed to detect, or possibly even correct for, misalignment between the transmission scan and the emission scan relies on acquiring a second, very short transmission scan either during the emission scan or immediately after it. This second transmission scan will be too noisy (and possibly too contaminated with emission data) to use for attenuation correction. However, it has been shown that when such short emission scans are reconstructed they may contain sufficient data to permit alignment between themselves and the previous (good-quality) transmission scan. This technique can be used as a quality control check to detect potential misalignment or (if it is certain that motion occurred between the two scans) to realign the high-quality transmission scan to match the emission data (presumed to align with the short transmission scan). This scheme has been tested for PET[52] but not as yet for SPECT.

Noise in Attenuation-Corrected Images. The attenuation data measured from the PET or SPECT transmission scanning are used to correct the emission data. Unfortunately, the measured attenuation data often contain considerable statistical noise. This statistical noise is a result of the very small number of photons that are able to penetrate the body during the transmission scan. This is especially true for large subjects. The statistical fluctuations in the attenuation data produce statistical fluctuations in the corrected emission images. Transmission correction, then, produces more accurate but more noisy images. Several methods have been proposed to reduce this additional noise in PET, and similar schemes are being considered for SPECT.

The simplest way to reduce noise in the transmission data is to smooth it (although care must be taken not to smooth along the angular direction).[53] Smoothing the attenuation correction factors does indeed reduce the noise in the resulting corrected PET emission data. Unfortunately, such smoothing can also introduce artifacts in the reconstructed emission images.[54] These artifacts are caused by a mismatch between the resolution present in the emission data and the resolution in the transmission data. The effect is significant only at the borders between low and high attenuating media, such as occurs at the boundary between the myocardial free wall and the lung. Smoothing the attenuation data blurs the low attenuation correction factors associated with the lungs into regions occupied by the myocardium in the emission data. This can cause artificial reduction of counts in the free wall compared with the septum.

Alternative attenuation noise reduction methods, based on image segmentation, have been proposed.[55-57] These methods would completely eliminate the noise in the attenuation correction factors. The array of attenuation correction factors obtained from a PET blank and transmission scan is first itself reconstructed to form an image. This image is an image of the attenuation coefficients of the body. It has been suggested that the attenuation image be segmented into three regions (air, soft tissue, and lung) and each of these three regions be replaced by a constant (and therefore noiseless) value of attenuation coefficient. This segmented attenuation image can then be used to calculate new, noise-free attenuation correction factors.[55-57] There are several potential problems with this method. First, it is probably not true that regions such as the lungs are really uniform in attenuation value. Second, it is not clear how accurately one must be able to define the lung/soft tissue borders nor is it clear what uniform attenuation

value should be used, especially in the lungs. Finally, the effects of resolution mismatch referred to above also must be accounted for in segmented images (which have "perfect" resolution). Despite these difficulties, preliminary results with PET[54-57] indicate that the method may hold promise for PET cardiac imaging, and the same may be true for SPECT.

It has also been proposed to use CT or nuclear magnetic resonance images to compute attenuation maps that can then be used to correct PET or SPECT emission data. Scanners have even been described that combine instrumentation to acquire both CT and PET data simultaneously.[58] Except when such special instrumentation is available, however, the problem remains as to how to align the CT or nuclear magnetic resonance image properly with the PET/SPECT image. This alignment problem presents a formidable challenge and as yet remains unsolved.

In conclusion, attenuation correction can be performed easily and accurately in PET, and techniques are rapidly evolving to reduce the problems associated with noise, patient motion, and postinjection measurements. Methods for performing attenuation correction for SPECT are developing rapidly. At least one commercial implementation is available, and others are likely to follow soon. Much additional work is needed to assess the clinical efficacy of the correction schemes, but there is the potential for a reduction in the variability of Tl and sestamibi normal databases, with a concomitant improvement in sensitivity and specificity. The ability to perform quantitatively accurate SPECT imaging is sure to benefit other aspects of cardiac imaging as well.

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