Scatter Correction in 3D Positron Emission Tomography

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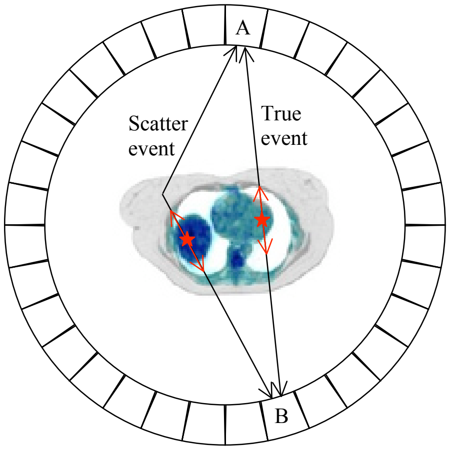


What is Scatter?

When a positron and electron annihilate in matter, they nearly always emit their energy in the form of two identical 511 keV photons traveling in nearly opposite directions. The rate of detection of such coincident pairs of photons should therefore be proportional to the total amount of positron emission activity along the "line of response" (LOR) connecting two detectors. Unfortunately, this simple picture is spoiled by Compton scattering. A 511 keV photon traveling through about 7 cm of water (or human tissue) has a 50 % chance of being scattered by an electron. For such "scatter" events the annihilation point need no longer lie along the LOR connecting the detectors (see figure), and consequently the simple relation between count rate and line integral is lost. In typical clinical scans 40–60% of the data may correspond to such scattered photon events.

Why is scatter a problem?

The problem with scatter is that tomographic reconstruction algorithms are based on the assumption that the data represent line integrals through the emitter activity in the patient. Scatter events are not consistent with this assumption, and are thus misplaced when included in the PET reconstruction. This can adversely affect final image quality and lead to image artifacts. Some examples are shown in the images to the right of a 135 kg female patient (scatter fraction 53%). The main point to realize is that an image uncorrected for scatter is not quantitatively accurate, and thus quantitative measures such as the standardized uptake value (SUV) may be incorrect.

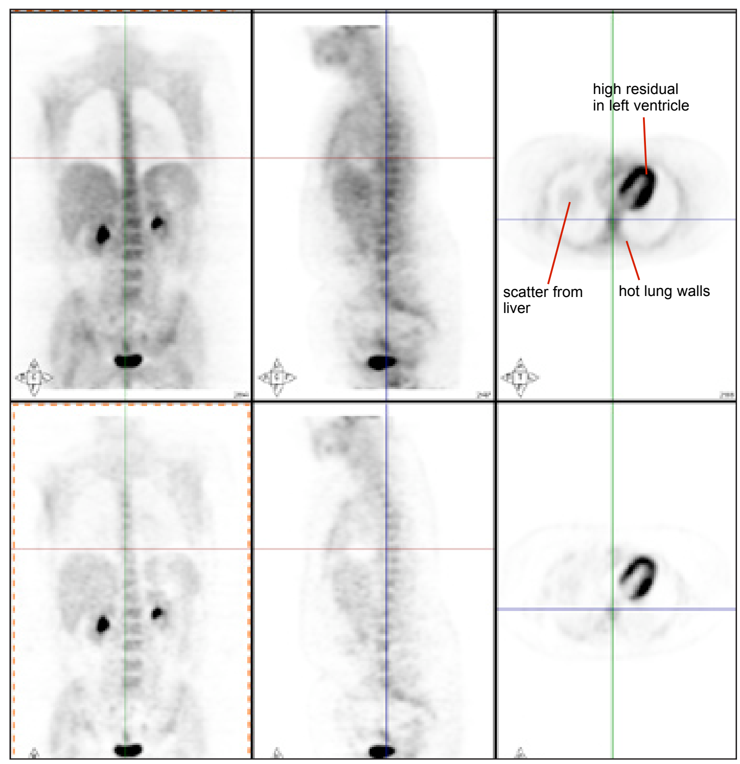
How can scatter be corrected?

Currently, all PET tomographic reconstruction algorithms treat the scatter as a "nuisance" component of the data, similar to random coincidences, that must be corrected in order to yield compatible data. Scatter estimation and correction in PET is a well-studied problem that has generated a large body of literature over the past 20 years. In general, approaches to scatter compensation may be classified as either empirical or theoretical.

scatter. Simple analytical models are common: for example, the use of Gaussian functions to fit to the scatter distribution in brain studies, or the use of a scattering "kernel" function to correct the sinogram via convolution-subtraction or deconvolution. This latter technique is in fact the standard approach used in 2D PET. The problem with these simple models is that they are not strictly consistent with the physics of scattering and can therefore be inaccurate, particularly in regions with significant attenuation structure – such as the thorax scanned with arms down.

Uncorrected

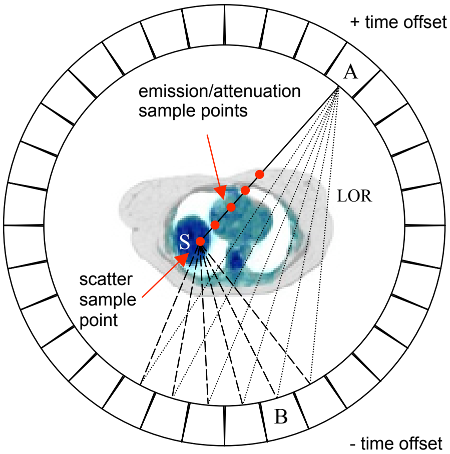
Scatter corrected



Empirical approaches involve the measurement of auxiliary data such as coincidence rates at photon energies below 511 keV, with the idea being that these must be scattered photons. Such techniques can potentially compensate for scatter originating outside the field of view of the scanner, but suffer from increased data processing requirements, increased noise, and the fact that the scatter in the auxiliary data may not have the same spatial distribution as the scatter in the primary data.

Most clinical scatter correction algorithms are based on theoretical models of the

Arguably the most accurate approach to scatter estimation is to compute it from the fundamental physics of the Compton scattering process. In research environments, Monte Carlo simulation has been widely used for this purpose. This technique allows the most complete representation of all physical effects, but because of the way such a calculation must be organized, together with the "noise" resulting from the essential randomness involved, this algorithm is not currently fast enough for general clinical use.

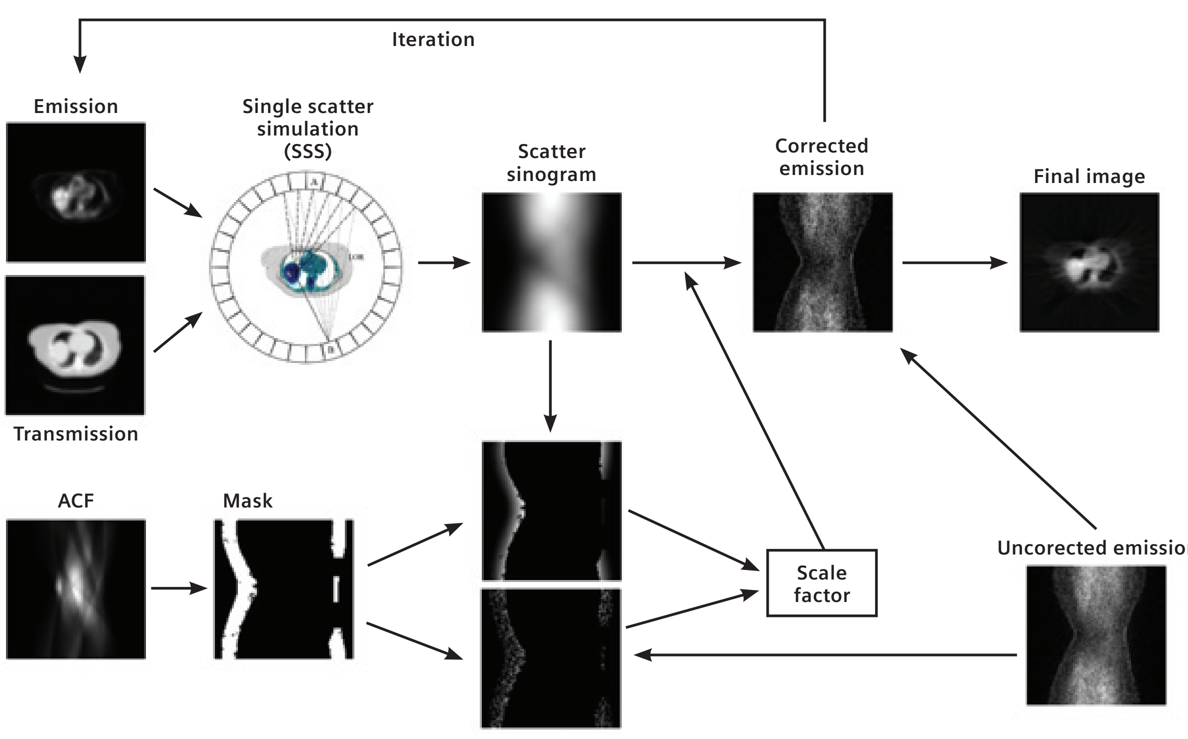


Single scatter simulation

It has long been recognized that if one neglected multiple scattering, which is a small component (about 20%) of the total scatter, and simulated only single scattering, i.e., events in which only one of the annihilation photons has been scattered once, then the scatter calculation could be performed much more efficiently than with Monte Carlo, and would become clinically practical. Neglect of multiple scatter is justified because it usually has little effect on the shape of the spatial distribution of the scatter interior to the body. Algorithms based on the single scatter approximation are in fact now the dominant scatter correction techniques for quantitative clinical 3D PET.

In the single scatter approximation the scatter coincidence rate in the sinogram LOR defined by a detector pair (such as AB) can be expressed mathematically as the volume integral of a scattering kernel over the body, represented by preliminary emission and transmission images. The accuracy and computational efficiency of the resulting scatter correction depends a great deal on how this numerical integration is done, and there have been several different implementations. The single scatter simulation (SSS) algorithm we have developed employs a unique, intuitive sampling technique organized as a summation over sample scattering points. This implementation is especially efficient because it reuses the computed ray sums through the object (such as AS) to compute scatter contributions to multiple LORs, as illustrated. Additional savings come from the fact that because the scatter distribution is typically quite smooth, the calculation may be appropriately coarse-grained without loss of accuracy. Accuracy is further improved by iterating the scatter calculation. To account for scatter originating from outside the field of view of the scanner, a sophisticated algorithm is used to scale the computed scatter sinogram to the emission data lying outside the patient, as determined from the attenuation correction factor sinogram. In PET•CT systems, when CT-based attenuation correction is used, care must be taken to exclude regions in which the CT image has been truncated, since these may actually contain LORs passing through the patient. With these and other refinements, the SSS scatter calculation typically requires only 20 seconds or less per patient bed position. The dataflow for the SSS algorithm is illustrated below.

The accuracy achieved by the SSS algorithm in estimating the scatter component of clinical emission data is remarkable – perhaps better than one might have expected from the simple physical model. One must look at a wide range of studies to fully appreciate its robustness, but the example below is typical. It shows measured emission and computed scatter sinogram projection profiles at various angles around a patient's abdomen. Also shown is the profile of the mask used for scaling. It's clear that neither a Gaussian fit nor a convolution-subtraction model would give an adequate estimate of the scatter in this case. It is important to reiterate that the SSS algorithm achieves this accuracy because it computes the scatter sinogram specifically for each individual patient scan from the fundamental physics. To our knowledge, no other clinical scatter correction algorithm provides this level of accuracy.

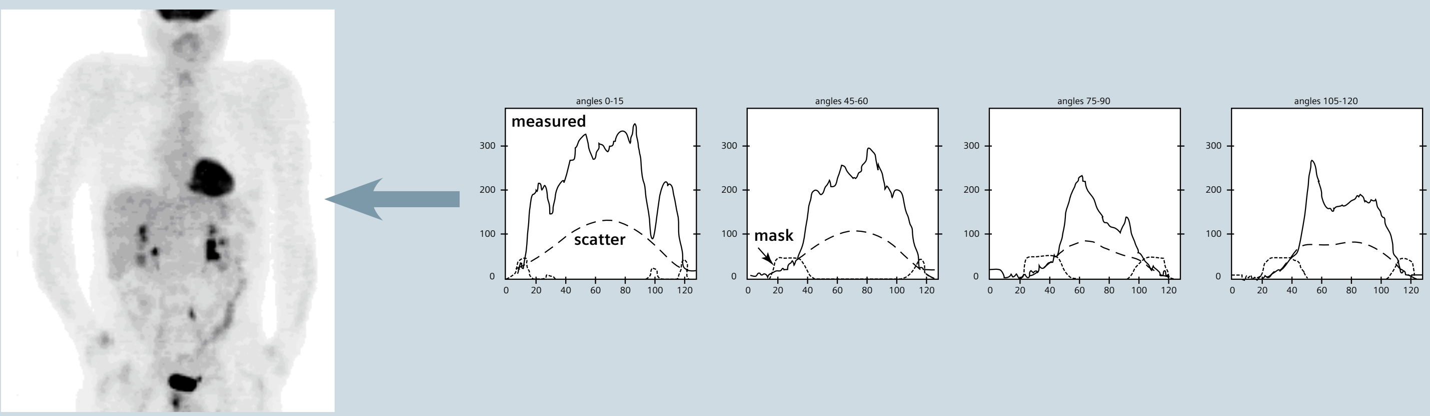


Time-of-flight SSS

With the advent of new, faster, high efficiency detector materials such as LSO, there has recently been renewed interest in the technique of time-of-flight (TOF) PET for clinical applications, due to its effectiveness in reducing image noise. TOF PET offers a new challenge for scatter correction since the scatter contribution to a specific measured time-offset bin for an LOR now depends not just on the trajectory of the photons, but also on the position of the annihilation point along this trajectory. The different time offset bins sample the object differently, and thus the spatial distribution of the scatter varies with the time offset. In other words, TOF scatter is significantly different from non-TOF scatter. Thus the scatter calculation must accurately account for the TOF physics or image quality may be adversely affected. The SSS algorithm has been adapted to TOF by incorporating a position-dependant TOF detection probability function in the calculation of line integrals along scattered photon trajectories. With this, TOF SSS has proven to be of comparable accuracy to the original non-TOF version in both phantom and human studies.

Summary

Accurate scatter correction is essential for quantitative PET: If we are going to throw away half the measured data before reconstruction, it had better be the right half! Single scatter based computational approaches, and in particular the SSS algorithm implemented in the Biograph, called True C, have proven to offer the best balance of speed, accuracy and robustness for a scatter correction that is precisely tailored to every individual patient scan. As PET moves into the future with TOF capability, an extension of SSS incorporating the TOF physics and the TOF SSS algorithm provides a practical solution for scatter correction for quantitative clinical reconstruction.



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