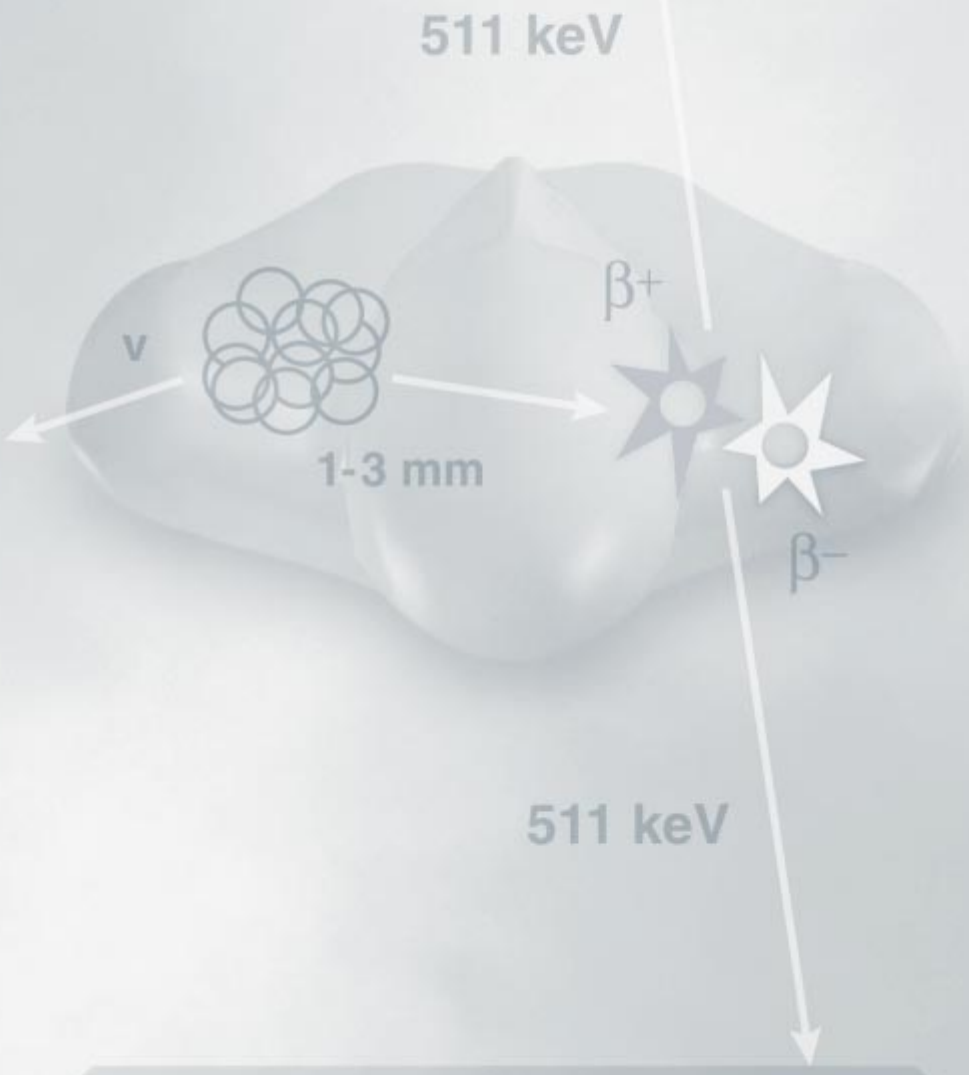
Positron
Emission Tomography Imaging with the Millennium VG



GE Medical Systems

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Introduction

Until recently, gamma cameras have performed only single photon studies. These studies have included both planar and tomographic acquisitions. The imaging of 511 keV photons has been primarily performed with dedicated PET scanners. These devices provide whole body tomographic acquisitions using radionuclides which have different properties. Thus, a different detector technology was designed specifically for the detection of positrons from the technology used on a traditional gamma camera.

Despite significant differences between Single Photon Emission Computed Tomography (SPECT) and dedicated PET, there have been attempts by several manufacturers to provide products that would fill the gaps between these two imaging techniques. Several options have been considered to achieve this, including designs for a dual-purpose SPECT/PET gamma camera. Newly designed collimation, thicker NaI crystals, and high performance detector electronics have allowed 511 keV photons to be detected by the gamma camera.

This paper provides an overview of PET technology applied to Dual Head Coincidence Imaging (DHCI). The discussion focuses on the intrinsic technology limitations of positron detection with the gamma camera and the mechanisms designed on the Millennium VG to improve detection capabilities.

PET Imaging on Gamma Cameras

PET Overview

PET is a medical imaging technology that generates high resolution images of human physiology. Unlike traditional nuclear medicine that relies on single photon emitting isotopes, PET uses radionuclides (tracers) like carbon, oxygen, nitrogen, fluorine and rubidium, which are the basic elements of (or used as analogs to) biological substances. PET produces images of physiological function, such as glucose metabolism, blood flow, organ perfusion, receptor-ligand binding rates, and oxygen utilization. With PET, normal and abnormal biological functions of cells and organs can be determined.

Coincidence Detection on Gamma Cameras

When PET technology is applied to the gamma camera, traditional nuclear medicine can take advantage of the positrons emitted from biological tracers. The clinical value of 511 keV imaging relies on the properties of these tracers and is dependent upon the ability of the gamma camera to detect the high energy photons. Positron imaging is currently being performed by both 511 keV SPECT imaging and coincidence detection methods.

511 keV SPECT imaging involves the use of detectors that are fitted with a 511 keV collimator capable of resolving these high energy photons. A clinical application for the use of this technique is for myocardial viability. In this case, only single 511 keV photons are detected in a cardiac tomographic acquisition.

The CoDe5 option transforms the Millennium VG into a dual-purpose system that can detect either single photons or positron emissions. Routinely, only a collimator change is required for either 511 keV SPECT imaging or coincidence detection imaging. The detection efficiency is provided by a 5/8" crystal and the use of photopeak-Compton window operating modes.

Radiopharmaceuticals

A broad spectrum of radiopharmaceuticals is used in PET imaging. In most cases, these PET tracer molecules retain the biological properties of unlabeled molecules. Carbon (^{11}C), nitrogen (^{13}N), and oxygen (^{15}O) are present in organic molecules, while other atoms can be substituted or used as analogs. For example, fluorine (^{18}F) can be substituted for hydrogen. Typically, radionuclides used in positron imaging have very short half-lives ranging from two minutes to two hours (Table 1).

Because of their short half-lives, ^{11}C, ^{13}N, and ^{15}O, can only be used when a cyclotron is installed near the imaging device. Rubidium, which also has a short half-life, is generator produced. In addition, the extremely fast decay of these radionuclides makes them less suitable for low count rate devices. However, ^{18}F is more flexible because it has a two hour half-life and can be distributed from a central radiopharmacy or institution having a cyclotron.

Although many types of positron emitting radionuclides have been used to label hundreds of molecules to study basic physiology and disease processes, one molecule has been used far more than any other. This molecule is ^{18}F-FluoroDeoxyGlucose (FDG) and is used in the majority of clinical PET procedures. FDG has a broad clinical application because every cell uses glucose as fuel. For the non-invasive evaluation of cardiac viability and lesion detection, FDG imaging on a dedicated, whole body PET scanner is considered the diagnostic "gold standard."

Radionuclide	Half-life (minutes)	Use	For
Carbon-11	20.5	Amino-acids	Clinical Research
Nitrogen-13	10	Ammonia (NH_3)	Cardiology
Oxygen-15	2.1	H_2O, CO, CO_2	Clinical Research
Fluorine-18	110	FDG and F-dopamine	Oncology, Cardiology and Neurology
Rubidium-82	1.3	Potassium analog	Cardiology

Table 1: PET Isotopes Sorted by Atomic Weight

Clinical Applications

Due to its physical and biological properties, FDG has broad applications and is showing some clinical promise when using gamma cameras in the areas of neurology, cardiology, and oncology.

The main tracer used in neurology is FDG. The brain is easier to image than most organs, mainly because of its smaller size as compared to a torso or a whole body acquisition.

Recent technological changes in gamma cameras have made them capable of improved 511 keV photon detection while using SPECT technology. New applications in the cardiology field have been opened by combining information from traditional nuclear medicine radiopharmaceuticals like 99mTc-Sestamibi for cardiac perfusion with FDG myocardial metabolism. Acquisitions for both Sestamibi and FDG are done in SPECT mode. The 511 keV photons from the FDG are not detected in coincidence, but are acquired by using high energy collimators similar to a traditional nuclear medicine procedure.

Malignant tumors accumulate FDG proportional to their high proliferative rates. Benign tumors usually have lower uptake of FDG, so they can be potentially differentiated from malignant tumors. Coincidence detection can be used to evaluate the presence of malignancy in patients with known or suspected metastatic disease (Figure 1).

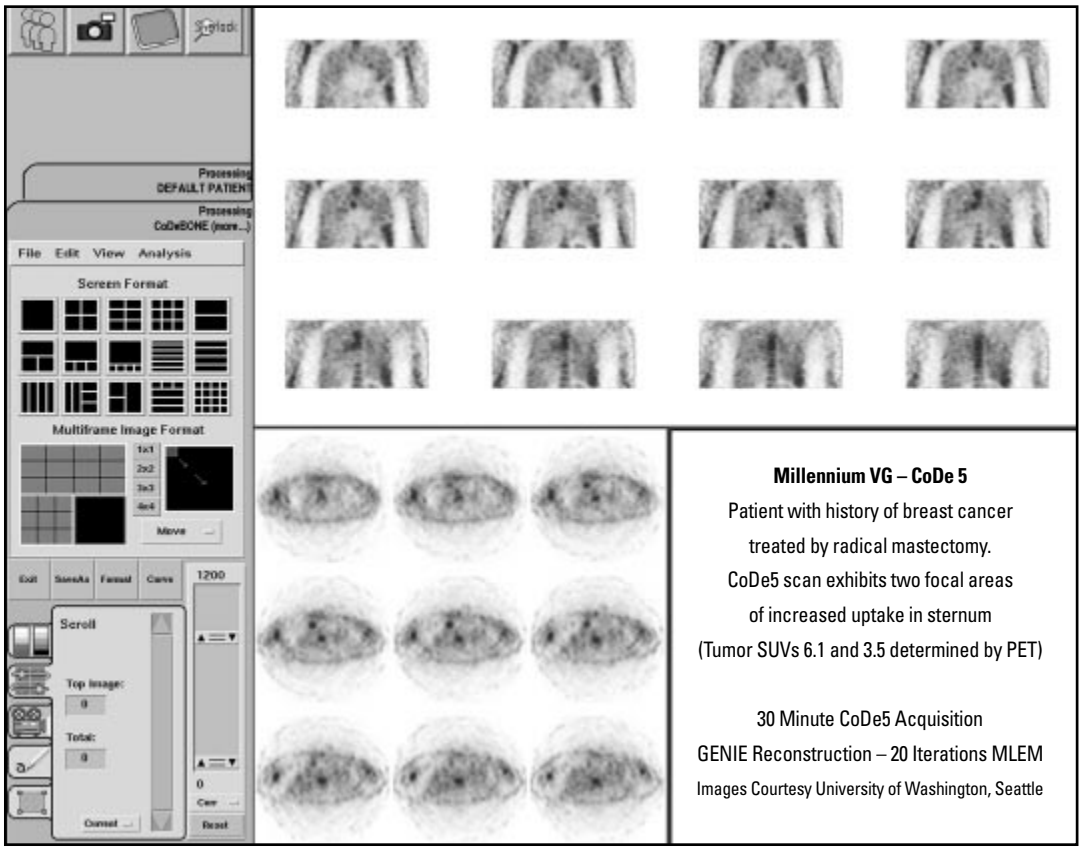


Figure 1: FDG Use in Oncology

PET/DHCI Physics

The aim of this section is to provide a brief overview of PET physics and discuss some of the issues regarding coincidence imaging on a gamma camera. Also, this section provides background information for understanding how the Millennium VG provides solutions to these issues.

Positron Emission

When a positron emitter such as ^{18}F decays, it emits a positron (the antimatter particle to the electron). Depending on the energy of the emitted positron, the positron will travel a few millimeters before nearly coming to rest and annihilating with an electron. The distance between the emission and annihilation of the positron plus the photon non-colinearity cause an uncertainty in positioning the event. The annihilation produces two 511 keV photons, traveling in nearly opposite directions (180 ± 0.6°) as seen in Figure 2.

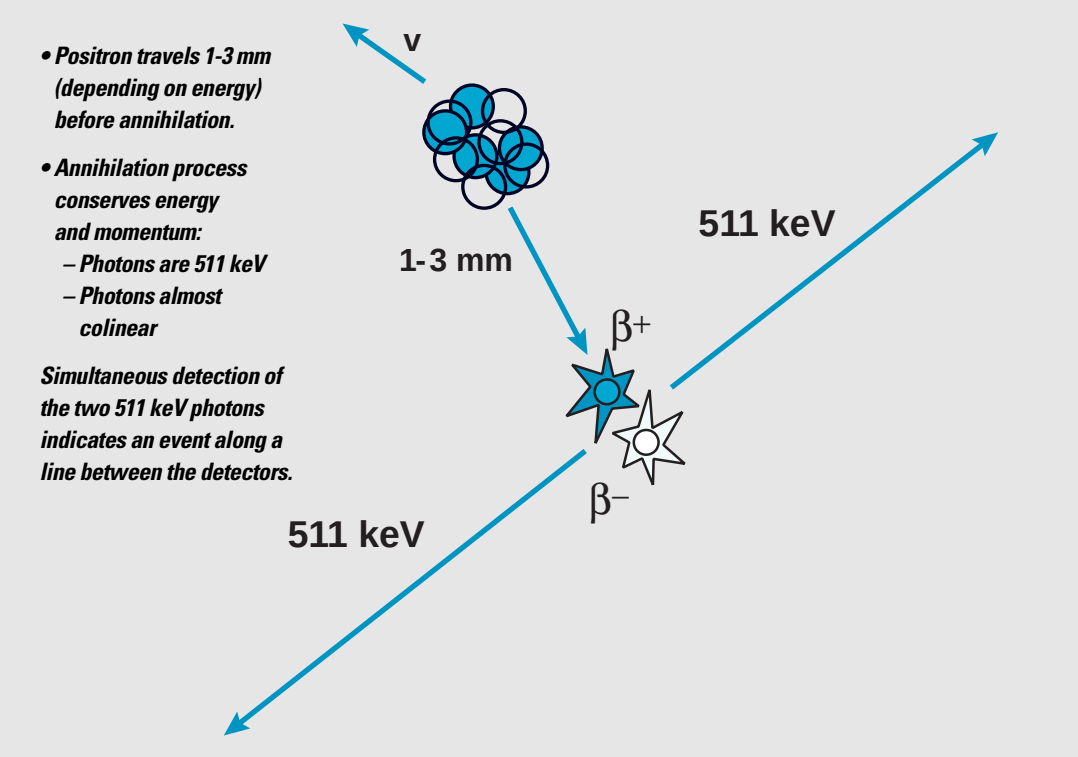
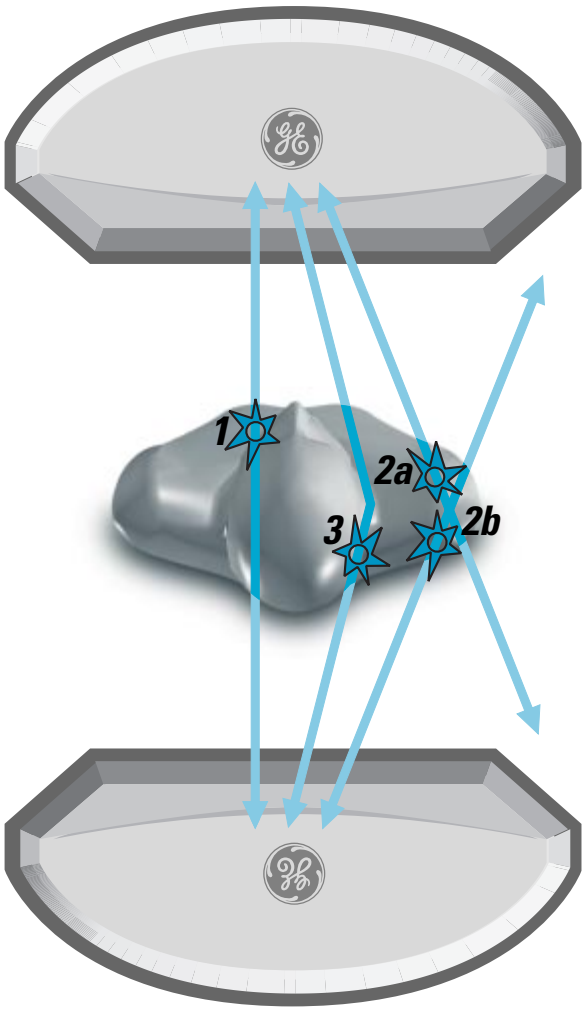


Figure 2: Positron Emission (Example of ^{18}F)

A PET system exploits these properties to "electronically collimate" the photon pair and determine the path along which the annihilation occurred. Since no physical collimator is required for event localization, it is this concept of "electronic collimation" that provides the modern PET system with a sensitivity and resolution advantage over single photon imaging systems.



1 - Detected True Coincidence Event
2 - Random Coincidence Events (2a, 2b)
3 - Scattered Coincidence Event

Figure 3: Types of Coincidence Events

Coincidence Detection

The concept of coincidence detection is quite simple: two photons are found to be "coincident" or "in coincidence" if both photons are detected by the system within a specified time interval known as the "coincidence window" (typically 10-20 ns). All events or Prompts (P) that are found in coincidence are either Trues (T), Randoms (R) or Scatter (S). These events are related by the formula:

$$P = T + R + S$$

Depending on the crystal type and thickness used in PET imaging, a portion of these events will be lost due to the lack of sensitivity and dead time.

The Trues represent the good data. As described later, the Randoms can be estimated and corrected, and the Scatter can be rejected using graded absorbers. The location of the annihilation event (hence the location of the positron emitter) is assumed to be along the path between the two detectors (Figure 3, Event 1).

This simple concept becomes complicated in practice because not all events found in coincidence are true coincidence events (Trues) as described.

Randoms

One such complication occurs when two photons from separate annihilation events are found in coincidence due to the finite size of the coincidence window (Figure 3, Events 2a, 2b). Simple probability provides that given Singles rates S_1 and S_2 for a pair of detectors and a coincidence window of width 2τ, events will be found in coincidence due to random occurrence with a rate:

$$R = 2\tau S_1 S_2$$

These random coincidences naturally occur within the coincidence window. Noting that the detector's Singles rate is proportional to the imaged activity, it follows that the Randoms rate is proportional to the square of the activity. This contrasts the Trues rate, which is only proportional to the activity. Randoms become a limiting factor at higher activities. Minimizing the coincidence window is important as well as any optimization that increases the Trues to Singles ratio. If the Singles rates for each detector pair are measured, Randoms can be estimated either directly by the formula or by a second delayed coincidence window.

Scatter

Another complication is caused by Scatter. One or both of the photons from a single annihilation event may scatter in the object of interest. This changes the direction and energy of the photon such that the position information of the event is lost (Figure 3, Event 3). These scattered events constitute a significant fraction of the events detected by the system (10 - 20% of total counts in a "2D system" and 40 - 60% of total counts in a "3D system"). The magnitude of the Scatter is dependent on the distribution of activity (inside and outside the field of view), the distribution and composition of the object of interest, and the design of the detector.

Image Count Considerations

The goal of a PET system is to collect a sufficient quantity of "good counts" (measured Trues versus Scatter and Randoms). Depending on the application, different detector attributes (such as resolution, sensitivity, count rate and the ability to reject Scatter and Randoms) will be of greater or lesser importance. For example, count rate is a critical factor for ^{15}O and ^{82}Rb flow studies in which a large bolus of activity (~ 50 mCi) is injected and imaged over a short period of time (< 1 min). This occurs in a flow study as the bolus travels through an organ such as the heart or the brain. In contrast, FDG whole body oncology studies do not stress the count rate capacity of most PET systems. These studies tend to be count limited. This can be due to the combined effect of clinically realistic imaging times, dose limitations, system sensitivity and attenuation. Instead, an insufficient number of counts are collected to support the available resolution of an imaging system.

Noise Equivalent Count Rate (NECR), which is defined as the square of the signal-to-noise ratio, expresses these factors as a single curve. Presuming there are accurate, unbiased corrections for Scatter, the NECR formula follows:

$$NECR = \frac{Signal^2}{\sigma^2} = \frac{T^2}{T+S+kR}$$

Where:
T = Trues
S = Scatter
R = Randoms
k accounts for the method of randoms correction:
k=1 for Randoms from Singles events
k=2 for unprocessed delayed event subtraction (real-time or from separate delayed sinogram)

Resolution

The resolution limit in positron imaging is largely determined by the intrinsic detector resolution. However, there is also a fundamental resolution limit that is a function of both the energy of the emitted positron and the photon non-colinearity. Colinearity refers to photon pairs that have an angle of 180 +/- 0.6˚ and causes a Full Width Half Maximum (FWHM) resolution loss of 1 to 2 mm for a 50 cm diameter detector. The combination of both effects leads to a fundamental resolution limit of 2 to 5 mm (Figure 4). This depends mostly on the imaged radionuclide and, to some extent, on the detector radius. For example, ^{18}F (Emax = 0.6 MeV) has better resolution (i.e., shorter distance before the positron annihilates into two photons) than ^{15}O (Emax = 2.1 MeV).

Several other factors will contribute to the final reconstructed image resolution as described later. For PET on gamma cameras, the reconstructed resolution is about 10 mm. For premium dedicated PET scanners, this resolution can reach 5 mm.

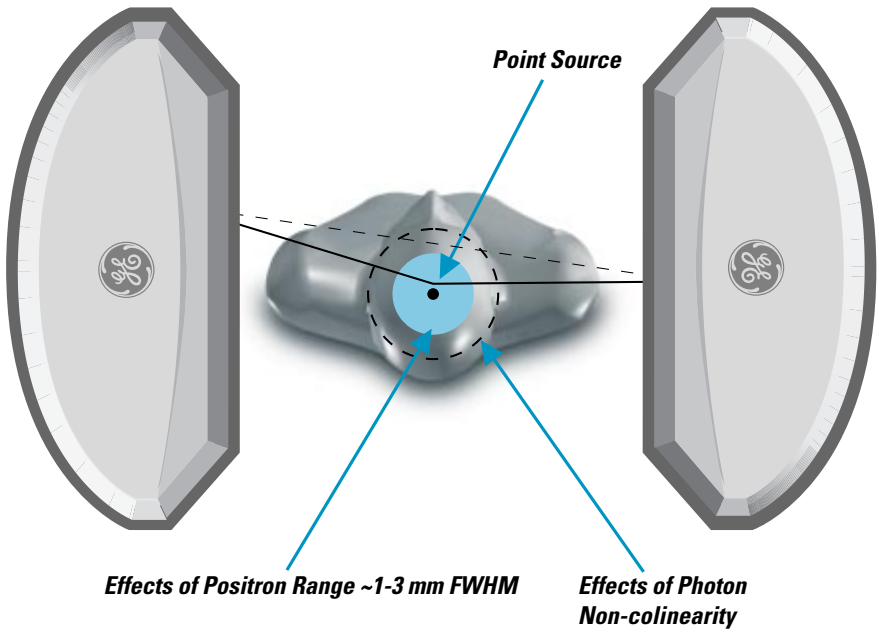


Figure 4: Theoretical Intrinsic Resolution Limits

Sensitivity

Sensitivity is ultimately determined by many factors like the detection efficiency of the crystal, system geometry and acquisition mode (2D or 3D). Since two photons must be detected to record one coincidence event, coincidence sensitivity is proportional to the square of the detector efficiency. High performance dedicated PET systems have evolved to use 3 cm thick BGO crystals. These systems provide a photopeak coincidence interaction probability of 85% at 511 keV. Compared to BGO, NaI has a lower "stopping power" and lower photo-fraction at 511 keV (See Table 2).

The standard 3/8" NaI crystal has less than a 30% probability of any interaction (photopeak or Compton) and about a 9% probability of a photopeak interaction at 511 keV. To obtain an equivalent efficiency to a 3 cm BGO detector, it is estimated that a 10 cm (4") thick NaI crystal would be required. While this is obviously impractical, even a modest increase from 3/8" to 5/8" can provide a 40% increase in the interaction probability and nearly a 90% increase for a photopeak interaction. In practice, 5/8" crystals provide a coincidence sensitivity that is three times greater than that of 3/8" crystals with minimal changes in SPECT resolution (See Table 3).

Measurements show that the increase in PET sensitivity is obtained with only a 0.4 mm degradation in intrinsic SPECT resolution. This results in a degradation from 7.4 mm to 7.6 mm in clinical resolution when using a high resolution collimator at 10 cm. Such a small degradation is assumed to be acceptable for SPECT imaging in order to support the significant gain in PET performance.

Photon Energy	3/8" NaI Dual Head	1/2" NaI Dual Head	5/8" NaI Dual Head	1" NaI Hexagonal Full Ring	3 cm BGO Full Ring
140 keV	85%	91%	94%	N/A	N/A
511 keV	9%	14%	17%	36%	92%
511 keV (in coincidence)	1%	2%	3%	13%	85%

Table 2: Overall Photopeak Efficiency Comparisons

NaI Crystal Thickness	3/8" (10 mm)	5/8" (16 mm)	1" (25 mm)
Intrinsic Resolution	3.5 mm	3.9 mm	5.2 mm
System Resolution (@140 keV, LEGP Collimator)	9.7 mm	9.8 mm	10.4 mm

Table 3: Crystal Thickness Versus SPECT Resolution

Uniformity

Uniformity needs to be considered for coincidence detection in both the axial and transaxial directions. The resolution at the edges of the axial field of view is critical to the performance of uniform scanning along the patient body. An increased number of Randoms events out of the field of view can cause a non-uniform distribution of counts at the edges of the crystal. This affects both 2D axial and transaxial uniformity.

Energy Spectrum

The NaI crystal is very efficient for low energy photons (for example 140 keV as in Figure 5).

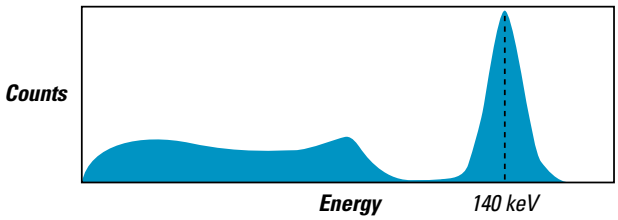


Figure 5: 140 keV Spectrum on a NaI(Tl) Crystal

Ideally, when a 511 keV photon interacts in the crystal, all the energy would be deposited and converted to light. That is not the case for NaI crystal based systems, even with an increase in crystal thickness. Approximately 80% of 511 keV photons that interact in a NaI crystal undergo Compton scatter. Many of the secondary photons escape the detector, leaving only a portion of the 511 keV incident energy in the detector (Figure 6).

This Compton scatter arises partly from 511 keV photons that collide in the crystal and lose only part of their energy. The remaining energy leaves the crystal in the

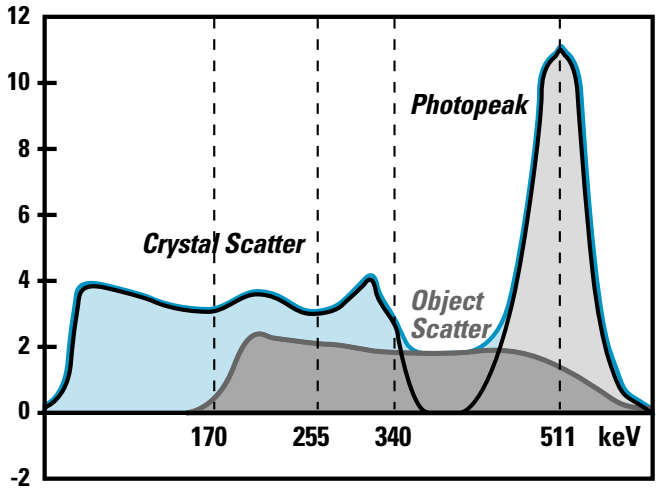


Figure 6: 511 keV Spectrum on a NaI(Tl) Crystal

form of another photon. The energy that is measured for Compton events is less than 340 keV, but the spatial information of the Compton scatter in the crystal is correct. Therefore, these Compton events are useful for improving the sensitivity with only a slight degradation in resolution. Some of these events can be captured by placing a second, or Compton, window below the primary photopeak window, and testing photons for coincidence from either window.

In addition to the crystal, Compton scatter occurs in the patient. Suppose we detect a photon that deposits 300 keV in the detector. Was this a photon that scattered in the patient or in the crystal? One cannot distinguish between crystal scatter and any object scatter from the patient body (Figure 6). The Trues/ Singles ratio and resolution could be improved if the most patient scatter is prevented from reaching the detector through the use of graded absorbers.

511 keV photon imaging in both PET scanners and gamma cameras follows similar physics fundamentals as described later. However, there are striking differences between gamma camera methods and dedicated PET imaging systems. The following table provides a comparison of different types of equipment used to detect 511 keV photons.

Typical Model Specification	Dual Head 5/8" NaI Collimated	Dual Head 5/8" NaI PET	Hexagonal Ring 1" NaI PET	Full Ring 30mm BGO PET
Scan Times in min.	30 - 40 (Torso)	120 (Whole Body)	60 (Whole Body)	30 (Whole Body)
2D mode Sensitivity in kcps/µCi/mL	N/A	25	N/A	200 - 220
3D mode Sensitivity in kcps/µCi/mL	N/A	275	400	900 - 1225
2D mode Noise Equivalent Count Rate (Peak) in kcps	N/A	2	N/A	97 - 159
3D mode Noise Equivalent Count Rate (Peak) in kcps	N/A	4	30	88 - 261
2D mode Scatter Fraction	N/A	24%	27%	10% - 16%
3D mode Scatter Fraction	N/A	37%	35%	35%
Intrinsic Resolution in mm	4 - 6	4 - 6	5.5	3.5 - 4
Clinical Resolution in mm	12	7 - 10	7 - 9	5 - 6
Target Clinical Application	Cardiac Viability	Oncology Neurology	Cardiac Viability Oncology Neurology	Cardiac Viability Oncology Neurology Clinical Research

Table 4: Typical Equipment Specifications for 511 keV Detection

Coincidence Detection on the Millennium VG

Design of Gamma Cameras For 511 keV Imaging

The primary difference between PET systems and gamma cameras that perform coincidence detection is the range of applications for which the system is targeted. PET systems are suitable for a wide range of PET applications, as they have specifications that excel in most attributes desired for a PET system. Today, applications for coincidence detection on a gamma camera are being investigated within one very important subset of all PET applications – namely those procedures based on the imaging of FDG. Coincidence options for gamma camera systems are evolving under the premise that the system is primarily a single photon imaging system (and hence must maintain single photon imaging utility). A simple transition to "PET mode" provides the gamma camera with an expanded set of applications.

PET imaging presents a new set of challenges for single photon imaging systems. There is a limit to the high energy performance of a system, due to a tradeoff between single photon performance for PET performance. Due to electronic collimation, the increase in sensitivity results in Singles rates of several million counts per second reaching the detector, stressing its count rate capability. Even with electronic collimation, the relative lack of sensitivity of the thin NaI crystal results in a system far less sensitive than a dedicated PET system. Sensitivity and count rate capability are of paramount importance in the design of a gamma camera coincidence option. As a result, one should be willing to trade some spatial and energy resolution in PET mode for increased sensitivity and count rate capability.

For gamma cameras to perform coincidence imaging, additional hardware components are required. On the Millennium VG, these hardware components are part of the basic configuration of the camera and include:

- Fast analog to digital converter for an accurate signal for improved energy and spatial resolution.
- A digital processor for pulse clipping, signal integration and short dead time.
- Computer control over gain, integration time (variable or fixed) and pulse clipping parameters for stability and flexibility.
- Dedicated hardware for pileup rejection and Singles loss measurement.
- Slip ring technology for revolving acquisitions.
- High energy graded absorbers and septa for 3D and 2D imaging.
- Data reconstruction as performed on a GENIE Processing & Review workstation.

Acquisition Geometry

Dual head coincidence imaging does not need the traditional collimators for nuclear medicine studies. While increasing the overall system sensitivity with an uncollimated detector, the problem of Scatter and Random counts (Singles) coming from outside the field of view of the camera needs to be addressed.

Depending on the acquisition mode, a reduction of Scatter and Random counts can be accomplished through the use of either graded absorbers or septa with graded absorbers. The Millennium VG uses High Energy Filter for 3D Acquisition Mode and High Energy Septa for 2D Acquisition Mode.

High Energy Filter

The High Energy Filter is used for three dimensional (3D) acquisitions. A 3D acquisition is done using a full set of photons regardless of the angle of incidence to the detector. The graded absorber is a "sandwich" of metal plates (in a specific order) that are placed between the crystal and the source of the Scatter. It is made of three layers, consisting of lead, tin, and copper with the lead layer being the closest to the patient (Figure 7). Lead preferentially absorbs the lower energy (scattered) photons from the patient rather than the high energy (unscattered) photons. The tin absorbs X-ray photons that are generated in the lead. The copper absorbs X-ray photons that are generated in the tin. The High Energy Filter reduces the photons scattered in the patient while letting through more than 80% of the events that were not scattered. This provides increased sensitivity for the Millennium VG when using various energy window combinations (as discussed earlier).

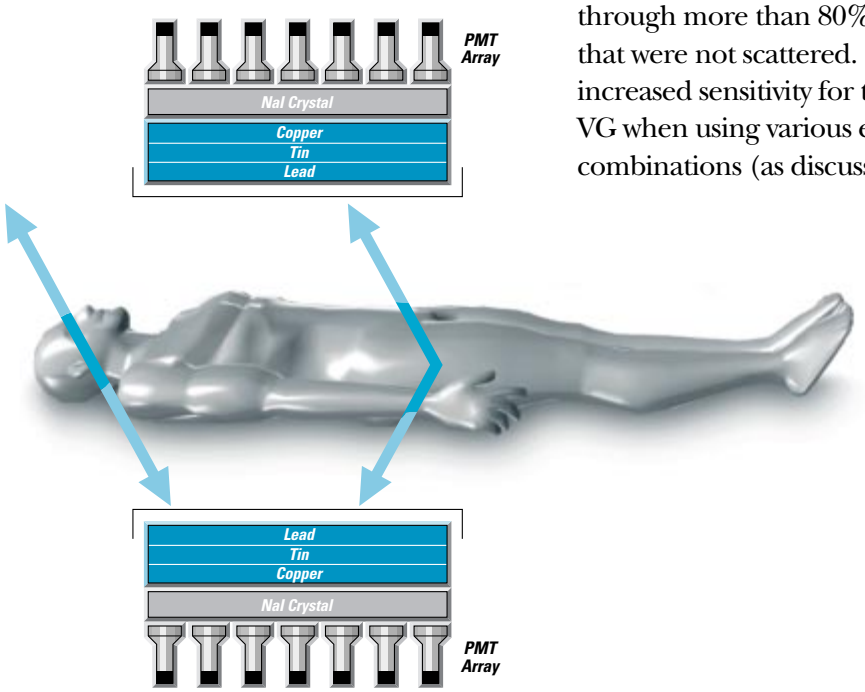


Figure 7: Millennium VG High Energy Filter Design

High Energy Septa

Two dimensional (2D) acquisitions are done using the High Energy Septa, collecting data from adjacent 2D planes. Septa (made of parallel slats) are introduced to axially reduce Randoms and Scatter originating from different planes or from out of the field of view (Figure 8A, 8B). Yet, when the septa are too close to each other, a lower sensitivity results (Figure 8A). Like the High Energy Filter, these septa contain the same three layers of lead, tin, and copper.

Although these septa reduce the sensitivity of the system to true photons, they will further reduce the sensitivity to single and scattered photons. This is especially true for those photons that originate from outside the axial field of view. Therefore, the percentage of true counts is increased, and the sensitivity profile in the axial direction becomes more uniform.

Without septa, the system is inherently a 3D acquisition system (Figure 8C). Three dimensional datasets are large, and full 3D reconstruction methods are computationally intensive. A reduction in data size can be obtained by a list mode acquisition and/or by an approximate binning of the data from 3D to a standard nuclear medicine 2D format.

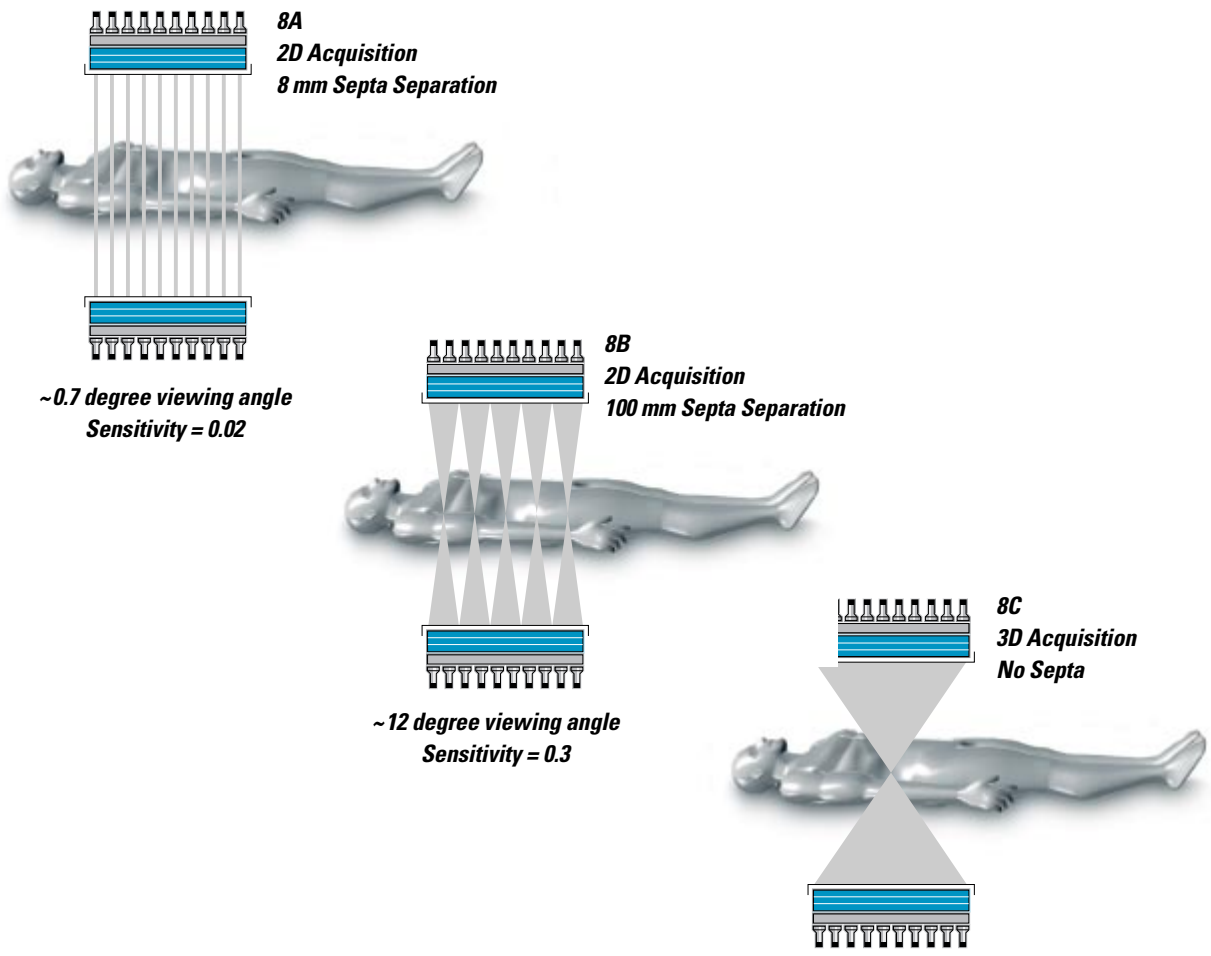


Figure 8: Inter-plane Septa on Dual Head Gamma Camera

Revolving Acquisitions

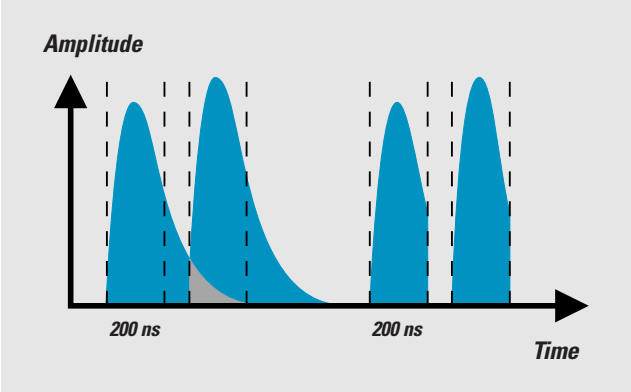
A coincidence acquisition on the Millennium VG is a multiple rotation acquisition, taking advantage of the unique slip-ring technology. The number of acquisition rotations can be determined by the user. The short decay time of FDG makes the revolving acquisition even more important. Gamma cameras, that acquire data in one single rotation, will need to make a larger correction for decay. As FDG decays, acquisition data obtained from latter angles will have lower activities than the initial angles. Since this type of acquisition requires the data to be decay corrected, it results in more noise amplification. However, with slip-ring technology, the Millennium VG camera acquires data in multiple rotations at faster sampling rates. In this way, the activity in the first view is not much different from the activity in the last view. Each of the acquisition rotations is then summed to have a more uniform activity requiring less decay correction. An added benefit of the slip-ring technology is that if the scan is aborted for patient motion, the acquired data can still be processed.

Signal Processing

Standard nuclear medicine systems are incapable of performing at high Singles count rates. Therefore, the detection electronics must be modified to provide for improved count rate capability. Since a 511 keV photon produces more than three times the light as a 140 keV photon, the same amount of light (and therefore equivalent position and energy information) can be collected in a shorter period.

Count rates can be extended by various techniques such as pulse clipping or pulse tail extrapolation. On the Millennium VG, pulse clipping is used to reduce signal pile-up and to increase the acceptance of true counts. To reduce pileup of signals, pulses are clipped to a duration of approximately 200 ns, which is less than the decay constant for NaI (Figure 9).

The Millennium VG CoDe5 system incorporates a variable integration time. At low count rates, the clipped signal is integrated to a specified time length of 300 ns. At higher count rates the integration time can have a minimum duration of 200 ns. This shortened integration time will produce slightly noisier position and energy signals, but permits a significant increase in useful count rate.



- **The amplitude of the second pulse is greater than the first pulse due to tail overlapping.**
- **Shorten pulses are clipped so that overlapping events can be separated.**

Figure 9: Pulse Clipping

Coincidence Windowing

It is important to note, however, that the Singles count rate in a coincidence system is not the only limitation to consider. Since Randoms increase as the square of the Singles rate, at some point the number of Randoms detected will equal or exceed the true events. With a Singles-to-coincidence ratio of 100:1, a Singles count rate of 1 Mcps per head and a coincidence window of 15 ns, a coincidence system will collect 10 kcps of true events (1 Mcps/100) and 15 kcps of Randoms [15 ns x $(1 \text{ Mcps})^2$]. Further increases in the Singles count rate capabilities are not worthwhile. While reducing the coincidence window width will produce modest improvements in true event capacity, the Singles-to-coincidence ratio must be reduced in order to dramatically improve the performance of the system. This enables the system to "count smarter, not harder." Randoms compensation is performed by using a delayed time window that is three times longer than the coincidence window (Figure 10). This makes the Randoms compensation more accurate by a factor of three in the counting statistics.

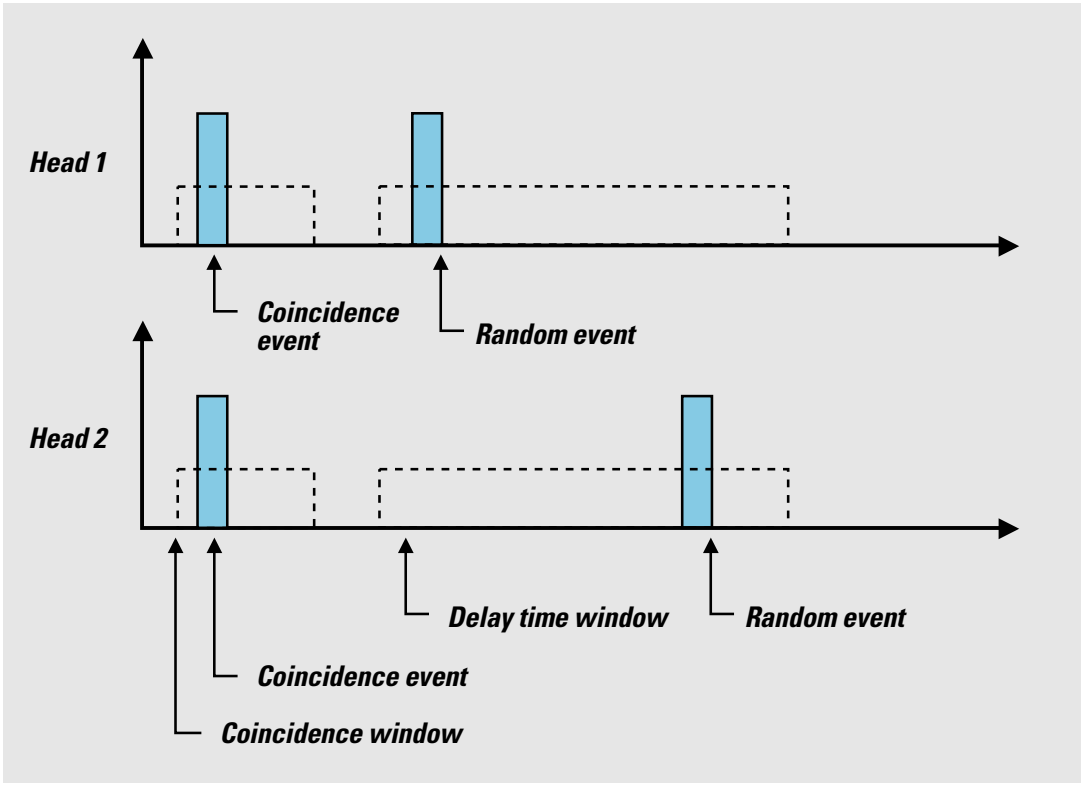


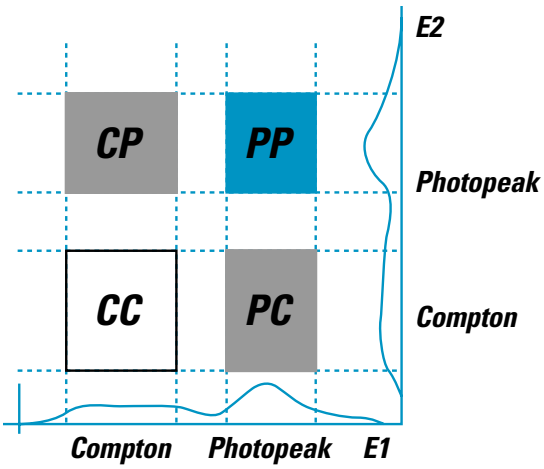
Figure 10: Randoms Compensation

Modes of Operation

With coincidence imaging, it is important to effectively use the Scatter that occurs in the crystal. This is accomplished by using various combinations of photopeak and Compton coincidence pairs.

There are three modes of operation available:

High Resolution Mode:
Only PP region is a valid coincidence.
Normal Mode:
PP, PC and CP are valid.
High Sensitivity Mode:
PP, PC, CP and CC valid. (High sensitivity mode is not recommended for clinical studies.)



PP - Coincidence Photopeak with Photopeak
PC - Coincidence Photopeak with Compton
CP - Coincidence Compton with Photopeak
CC - Coincidence Compton with Compton

Figure 11: Photopeak/Compton Imaging

Accepting photopeak-Compton coincidences in addition to the photopeak-photopeak events can roughly double system sensitivity. This may result in a net improvement in image quality. While the system can accept Compton-Compton coincidence pairs, this is generally not done because there may be a substantial loss of image contrast from accepting coincidences between two Compton window photons.

Data Preprocessing

In the Millennium VG, the system count rate is increased using a programmable (200 ns) digital pulse clip and variable (200-300 ns) integration time to address the needs for coincidence detection. Each system is tuned to provide a small coincidence window (11-13 ns typical value) to reduce Randoms. The system is capable of 2D list mode acquisitions using the High Energy Septa and 3D list mode acquisitions using the High Energy Filter. Online corrections are performed for energy, linearity, sensitivity and center of rotation.

Since the acquisition is done in list mode, the data is binned via single slice rebinning (SSRB) into projections for use with standard SPECT reconstruction techniques. With the Millennium VG, the user has the option of choosing the number of projections (60 or 90) in a 180˚ arc.

A log file, created by the rebinning utility, details the statistics of the acquisition, including the number of true coincidence events in the study. The list mode data is acquired into a Singles list (SLIST) that keeps the data separated for each head. Although the SLIST data can be processed directly, the normal procedure is to convert it into a coincidence list (CLIST) with user definable parameters, such as the acceptance angle. The list mode data is then binned (SSRB) into 2D sinograms for reconstruction.

Reconstruction

The rebinned data from the gamma camera can be reconstructed like any SPECT study, using filtered back projection or iterative reconstruction on a GENIE Processing & Review workstation in the case of the Millennium VG.

Iterative reconstruction appears to improve the quality of the image, which is limited by low statistics. Methods for iterative reconstruction can provide more quantitative imaging information and incorporate the effects of noise, counting statistics, missing data and attenuation.

A typical iterative algorithm uses the planar data projections and produces images according to the following steps (Figure 12):

1. An initial estimate of the transaxial distribution is created. This estimate may have the total acquired counts uniformly distributed or may be a closer guess from another reconstruction process such as filtered back projection.
2. Data estimates are forward projected from the current image estimate.
3. The difference between the data estimate and the raw data is computed. The resultant error image is used to update the image estimate.
4. Steps 2 and 3 are repeated a number of times, or iterations, until predefined termination criteria are met.

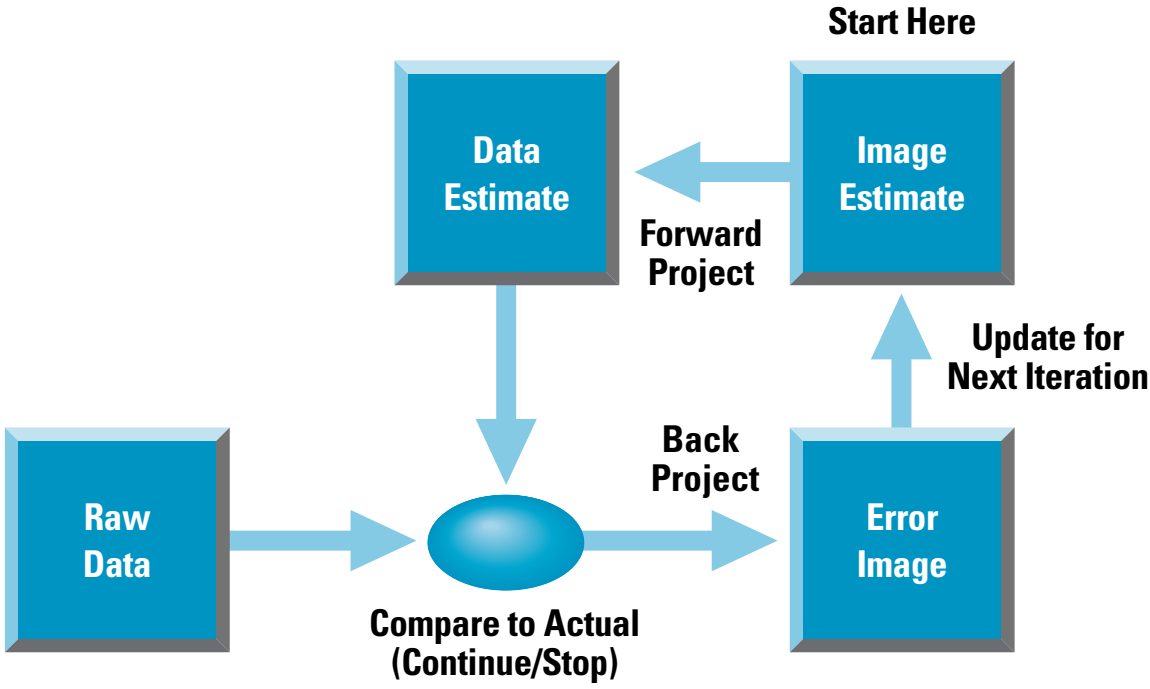


Figure 12: Iterative Reconstruction Flow Diagram

After the data has been reconstructed, emission post-processing involves the application of a filter. For iterative reconstruction, a three dimensional filter is applied to the emission transaxial slices following iterative reconstruction. If the post filter is changed, only the final post-processing step needs to be repeated. It is not necessary to repeat the entire iterative reconstruction process.

CoDe5 Summary Diagram

A summary flow diagram for high energy detection (CoDe5) on the Millennium VG is presented in the figure below.

The detectors provide a UFOV (Useful Field of View) of 540 x 400 mm and are mounted on a Slip-Ring gantry, allowing for continuous rotation. The radial head separation can reach 70 cm. Each detector features a High Energy Filter for 3D acquisition mode and High Energy Septa for 2D acquisition mode.

Signals from each PMT (photomultiplier tube) are converted by an ADC (Analog-to-Digital Converter) and then processed by an onboard DSP (Digital Signal Processor). This ensures control of pulse clipping and integration to achieve higher count rates and reduced pileup. The Coincidence Board receives energy signals from both detectors and generates signals from coincidence, Singles and pileup events. It has separate variable prompt and delay windows, pileup rejection, and multiple energy windows for photopeak-Compton imaging. Accepted signals are spatially positioned in a matrix by the Position Processor.

The Millennium VG Acquisition System makes it as easy to switch from SPECT to CoDe5 mode as changing collimators and selecting a different acquisition. Operation modes are selected as photopeak only or increased sensitivity, which includes Compton interactions. It has standard and customizable brain and whole body protocols with variable numbers of projections, matrix size, acceptance angle, zoom and speed of rotation. Single Slice Rebinning (SSRB) is performed for 2D acquisitions.

Finally, the projection data is sent via DICOM to the GENIE Processing & Review workstation to be reconstructed using either Filtered Back Projection or Iterative Reconstruction.

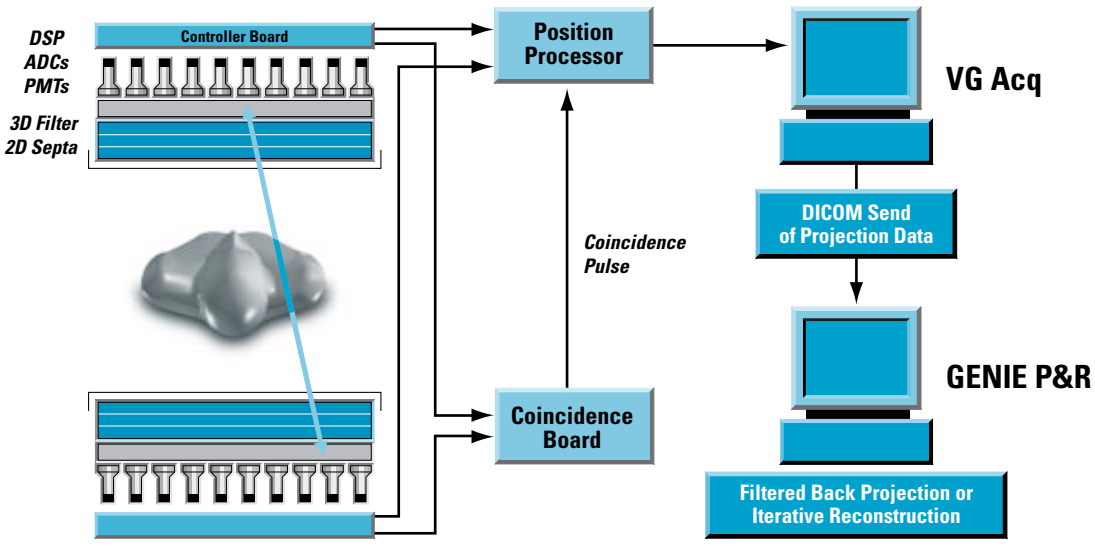


Figure 13: CoDe5 Summary Diagram

Summary

FDG whole body imaging is a very promising and potentially important clinical application. Most nuclear medicine manufacturers have responded to this need by adding the ability to image positron emitters to their dual head gamma camera systems. These systems are now being used in clinical settings.

For 511 keV imaging to be successful, several key factors need to be addressed including FDG availability, camera efficiency of high energy photons, procedure reimbursement, and training. Also, it is important for equipment manufacturers to develop dual-purpose "SPECT/PET" systems that can deliver:

- Both low and high energy photon detection
- Rational operating conditions for both patient and department
- Uncompromised SPECT performance
- An ability to upgrade existing gamma cameras to coincidence detection capabilities
- A sound financial investment on the asset

Over the next couple of years, it is likely that the clinical efficacy of coincidence detection on gamma cameras will be better defined. It is equally likely that significant improvements in the performance of these systems will also occur.

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