

Recent Developments in Multiple Detector SPECT

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INTRODUCTION

Multiple detector SPECT systems have higher sensitivity and the ability to collect complete data sets more quickly than the prevalent single head cameras. The major problems affecting quantification are scatter, attenuation and geometric collimator response. The scatter distribution in a projection image varies not only from patient to patient on the same system but from view to view on the same patient and from pixel to pixel in the same view. The scatter correction method is not based on simple filtering or deconvolution techniques. It does not rely on simple approximations of scatter content such as linear, polynomial or theoretical function filtering. It is not a simple subtraction of images from different energy windows which assume that the scatter fraction remains constant throughout the image area, when in fact it depends entirely on the geometry of the object being imaged. As the matter of fact the scatter content is not stationary and the analysis of the energy spectrum throughout and beyond the photopeak, is not simply at the endpoints. Attenuation artifacts are the major source of false positive scans and decreased specificity. Several SPECT manufacturers now include an attenuation correction for myocardial perfusion study. The scatter elimination by spectral acquisition memory extension is added for regional image contrast improvement and quantitative accuracy.

Scatter Elimination

As much as 30 to 40 percent of number of counts acquired in a patient study arrive at the detector after undergoing scatter interactions. Scattering changes the photon trajectory which subsequently leads to incorrect identification of the point of origin of the acquired photon. This directly translates into a reduction in both image contrast and detail.

An acquisition - based scatter removal technique which analyses the spectral distribution of scatter by measuring the energy spectrum of each pixel and removing

the scatter content. Simultaneous acquisition and process projection data using an integrated hardware and software to perform fine spatial and spectral sampling by utilizing up to 128 energy channels with spectral sampling of 1 keV per channel. It determines the scatter content pixel by pixel based on the naturally occurring variations of the measured energy spectrum. Fine spatial sampling of energy spectra helps visualize the rapid change of scatter content at object boundaries enhancing the effectiveness of edge detection in the corrected image. It is necessary to realistically quantify scatter content. For example,

1 keV shift due to scatter from 140 keV ^{99m}Tc photon, this indicates a scatter angle of 13 degree. If the scatter site is 10 cm from the detector, the spatial error of this photon is 2.2 cm. The scatter elimination results in the contrast improvement, detail enhancement and the increasing quantitative accuracy. Since the scatter is removed, measured attenuation of activity approaches true narrow beam performance.

Attenuation Correction By Transmission Information Observation Network With Detector Spread Function

In order to correct for attenuation and detector blurring in SPECT imaging, patient specific transmission projection data is acquired using a scanning line source system attached to a multiple detector system. The scanning line source system consists of a line source holder, electronic mask circuitry and a motion control subsystem. The line source holder covers the entire field of view of its corresponding detector without truncation caused by use of fan beam collimator. The line source is collimated with lead using a single slit design providing a focused stream of photons with minimum radiation exposure to the patient. ^{99m}Tc line source of activity 300 mCi is used and the patient is exposed less than 3 mR during 10 minutes transmission study.

The electronic mask circuitry creates a moving adjustable width electronic mask (electronic collimation) which helps eliminate scattered transmission photons. A digital encoder coupled to counter circuitry tracks the position of the source holder. Source position information from the encoder/counter is used to create a moving electronic mask which rejects photon that are not traveling in the path perpendicular

to the detector face, i.e. scattered photons.

The motion control subsystem is designed to move the source holder at a constant speed in the axial (y) direction across the opposite detector field of view. A high performance motor drive produce a speed constancy that yields a flood uniformity equivalent to the intrinsic uniformity of the camera. The motion control circuitry is synchronized with the gantry motion so that initiation of the line source motion is precisely coordinated with each ECT view. The speed of the line source motion can be scaled to accommodate a range of view times.

The combination of physical and electronic collimation provides a scatter free transmission image. The parallel beam geometry generates a truncation free transmission map.

The transmission projection data, along with the measured incident flux are used to generate an attenuation map. An iterative algorithm, which incorporates the transmission map and point spread function of the detector, is used to correct for photon attenuation and depth dependent detector spread function.

Motion Correction

Many studies described technique for detecting and correcting for patient motion. Some method⁽¹⁾ used frame to frame cross correlation functions of the summed profiles in the vertical and horizontal directions of the planar projection images. This is important for dynamic renal, gastric emptying studies where patient motion interferes with accurate time-activity curves. Other study⁽²⁾ uses diverging squares as automated methods for not only detecting but also correcting for patient

motion. The diverging squares algorithm is relatively insensitive to noise and requires no empirical parameters such as threshold values or noise statistics. One study uses fiducial markers to track and correct for whole body motion. Some manufacturer puts motion correction toolbox in the basic system to correct both planar and SPECT images.

FDG SPECT Studying

The imaging of positron emitters with standard nuclear medicine equipment is not a new practice. Many ^{18}F bone scans were obtained with the use of rectilinear scanner in the earlier days of radionuclide imaging. Minn⁽³⁾ et al have used planar acquisitions for FDG imaging in patients with head and neck tumors and have shown good preliminary success with this tech-

nique. Drane⁽⁴⁾ et al designed whole body SPECT system at the outset to also permit performance of 511 keV SPECT

Fluorine-18 labeled fluorodeoxyglucose (FDG) is found very useful to provide information for brain, heart and whole body studies with PET system. As a cost of a PET center is large and there are uncertainties regarding reimbursement and somewhat tenuous profitability make the establishment of a PET center a difficult financial proposition in USA. The use of ^{18}F FDG for myocardial viability is a low volume but steady demand. It is also used for clinical and research for cancer studies. Figure 1. shows the comparison of myocardial perfusion images between Tl SPECT (row 1 and 4), FDG SPECT (row 2 and 5) and FDG PET (row 3 and 6) of

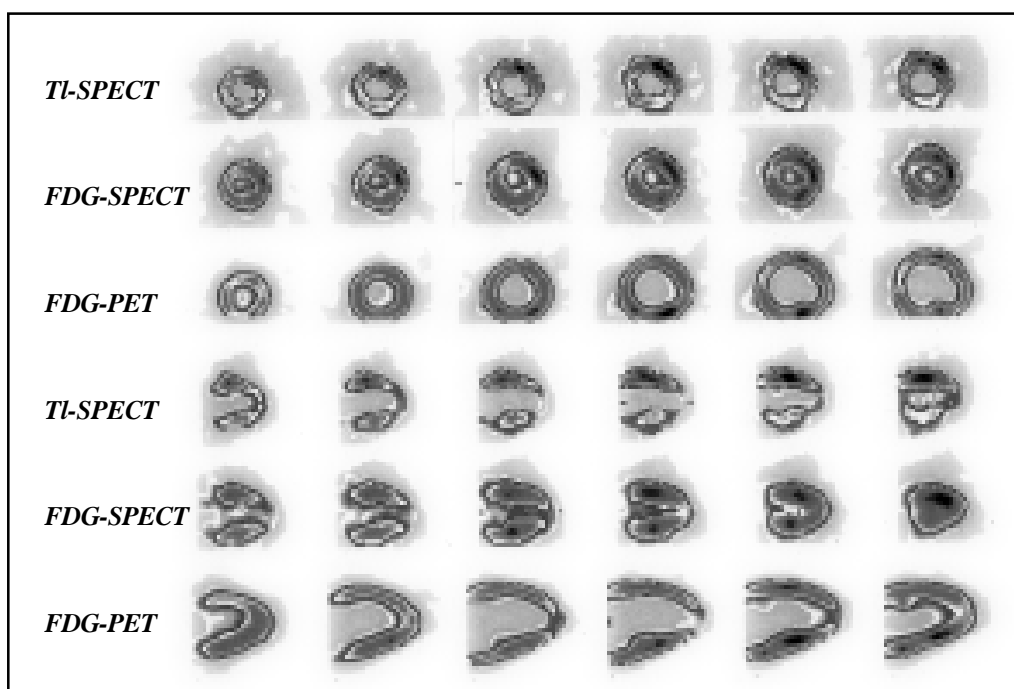


Fig. 1 SPECT and PET images of myocardial perfusion using ^{201}Tl and FDG.
Courtesy of Dr. Burt et al., VA Medical Center and Indiana University School of Medicine and JNM

short axis (row 1-3) and vertical long axis (row 4-6). The development of whole body SPECT for positron imaging becomes attractive alternatively to the other ways to obtain myocardial FDG imaging data and the oncologic uses of FDG imaging without a substantial investment in dedicated equipment.

New development on high energy collimator, capable of imaging in the 500 keV range or new coincidence detection capabilities to produce FDG SPECT are about one-third in assessing myocardial viability and two-third for oncology patients.

One advantage of FDG-SPECT is the ability to use it in conjunction with a single photon emitter for dual isotope imaging, ^{99m}Tc MIBI for myocardial perfusion and FDG for viability assessed in one study.

SPECT-High Energy Collimator Designed For Positron Imaging

Crystal thickness:

As the crystal thickness increases the efficiency of the detection of high energy photon increases but the spatial resolution for standard radiopharmaceuticals would decrease.

Side shielding:

When the detector shielding increases for high energy photons it would limit the ability of camera to image the cerebellum during close orbit around the skull at cerebral perfusion imaging with technetium imaging.

Collimator:

The high energy collimators were designed to optimize relative resolution versus sensitivity. Parallel hole high energy collimator for 400-500 keV photons was designed of about 450 lb or 210 kg weight, the collimator thickness is 7.81 cm, the hole diameter is 3.81 mm, lead wall thickness 1.71 mm and the septal penetration is 8 percent.

Gantry:

It must be designed to be able to support the unusually heavy collimators of at least 450 kg for dual collimators.

Detector performance:

Table 1 compares detector performance for ^{99m}Tc and ^{18}F and table 2 shows energy, spatial and reconstruction resolutions for ^{18}F -FDG.

Table 1 Comparison of detector performance between ^{99m}Tc and ^{18}F

	Energy window (%)	Planar sensitivity (cpm/ μCi)	Resolution at 10cm (mm.)	
			planar	tomography
^{99m}Tc	20	65	6.6	9.3
^{18}F	15	52	8.2	10.2

Table 2 Resolution of ^{18}F -FDG

Energy Resolution %	Spatial resolution at collimator face (mm)		Reconstruction resolution at 11 cm (mm)	
	UFOV	CFOV	FWHM	FWTM
8.96	1.92	1.87	10.7	22.7

FDG SPECT Clinical applications

FDG SPECT can be used for myocardial viability studies, brain studies, oncologic evaluation on head and neck malignancy, pancreatic cancers, breast cancer and miscellaneous. In the case of brain studies, FDG is used for the study of brain metabolism and the comparison is made for ^{201}Tl SPECT for evaluation of gliomas.

The performance of FDG SPECT to search for unknown primary tumor sites or to evaluate therapeutic intervention in patients with cancer is still considered research. FDG imaging has a potential role in the management of head and neck cancer patient. Its role is limited at the time of detection. If the primary tumor site is obvious, FDG imaging currently offers little advantage to CT for staging. If the primary tumor cannot be localized, however, FDG imaging becomes important. Its value in follow up after therapy increases. It has potential to differentiate radionecrosis from recurrent and to detect tumor prior to anatomic change, both areas clarify not well performed by CT. It is clear that granulation and inflammation tissue will occasionally take up FDG.

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