COUNT RATE AND SPATIAL RESOLUTION PERFORMANCE OF A 3-DIMENSIONAL DEDICATED POSITRON EMISSION TOMOGRAPHY (PET) SCANNER

By

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To my parents
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COUNT RATE AND SPATIAL RESOLUTION PERFORMANCE OF A 3-DIMENSIONAL DEDICATED POSITRON EMISSION TOMOGRAPHY (PET) SCANNER

By

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The count rate and spatial resolution performance of a state-of-the-art 3-dimensional dedicated positron emission tomograph at Shands Medical Plaza were assessed. A 70-cm-long polyethylene phantom and a line source were both utilized to better study the effect of activity from outside the field of view on the counting rate and spatial resolution of the scanner. A series of 21 tomographic images were acquired with a fluorine-18 radionuclide. The computed true, scatter, random, and noise equivalent count rate coincidences were plotted as functions of activity, as described by NEMA 2001 protocol.

The spatial resolution was studied by calculating the full width at half maximum (FWHM) of the fitted final sum projections. The FWHM decreases over time as activity decays, and is constant for the last 5 acquisitions with activities ranging from 0.22 mCi for the 17th acquisition to 0.08 mCi for the 21st acquisition. Moreover, the modulation transfer function was computed to completely describe the spatial resolution at varying...
activities. It was concluded that both the count rate and the spatial resolution were affected by the increase in activity, especially at high counts. It was found that the peak NEC rate occurred at 31.04 kcps at an activity of 5.876 mCi. The scatter fraction of the system was found to be about 40% indicating the disadvantageous high scatter coincidence events due to the contribution of scatter from out of the field of view. Finally, the peak NEC rate is a useful measure of the amount of activity that should be employed in a clinical patient scan. Also the trends of the true, scatter, and random count rate curves show the behavior of the scanner studied.
CHAPTER 1
INTRODUCTION

There is increasingly wide PET application for clinical diagnosis, due to the improvement in the diagnostic accuracy of this imaging modality through the development of new data acquisition and processing systems and the introduction of new positron-emitting radiopharmaceuticals. Positron emission tomography (PET) with fluorine-18 fluoro-2-deoxy-glucose ($^{18}$F-FDG), a glucose analog, is a unique biologic imaging tool that can be used to obtain functional information from the living human body. After administration of the $^{18}$F-FDG radiopharmaceutical to a patient, it takes part in physiologic processes. The unstable, neutron-deficient nuclide decays by emitting a positron, which will annihilate with a nearby electron to create two 511 keV photons that are emitted essentially back-to-back at an angle of 180 degrees. Detection of the two photons within a narrow time window is called a coincidence event, where the assumption is that the photons originate from a single annihilation occurring along the line-of-response defined by the direction of the photons.

The radioactive decay is continuously taking place and can be detected from the emitted gamma rays, allowing the monitoring of the distribution of the tracer concentration. The widely used $^{18}$F-FDG has an advantage of becoming trapped in the tissue and reaching a near equilibrium state approximately 45 to 60 minutes after injection, therefore keeping the tracer concentration almost constant with time.\(^1\) Oncology imaging using PET with $^{18}$F-FDG has proven to be effective in evaluating the extent of disease for several types of cancer.\(^2\) FDG-PET is useful for defining the degree
of malignancy and for differentiating recurrent tumors from necrosis after therapy in patients with brain tumors.³

Data acquisition is substantially different in PET than it is in planar nuclear medicine and single photon emission computed tomography (SPECT). In PET, annihilation coincidence detection (ACD) is used in lieu of absorptive collimation to determine the directionality of the detected photons. Because of this technique, the sensitivity, the rate at which the system detects counts per unit of activity, of PET is substantially higher than in single-photon nuclear imaging. ACD also avoids the degradation of spatial resolution with distance from the detector.⁴ There are two basic ways of collecting data in PET. These are referred to as 2-dimensional (2D) and 3-dimensional (3D) PET. Some PET scanners can acquire data in both 2D and 3D modes whereas others can only acquire data in 3D mode, like the dedicated PET scanner considered in this investigation. With 2D PET, thin rings (~ 1 mm thick) of lead or tungsten, known as septa, are often placed between the detector rings to reduce the interplane scatter. These septa are more like the antiscatter grids used in radiography. They do not provide spatial definition but simply reduce the amount of interplane scatter incorporated into the data acquisition. On the other hand, the absence of septa in 3D PET increases the sensitivity by a factor of 4-6 but also increases the scatter fraction as well as the random coincidences from activity that is out of the field of view.⁵ The increased sensitivity to true events is partially offset by the higher sensitivity to scattered and random coincidences. Consequently, the overall advantage of 2D versus 3D PET whole-body imaging is unclear.⁶
In the last 11 years, whole-body $^{18}$F-FDG studies have become the predominant type of PET study performed by most centers. Therefore, it is more relevant than before to measure the performance of the PET scanners under conditions that better represent whole-body studies. The phantom that was used for evaluating the performance of the scanner was a 19-cm-long phantom. The phantom’s volume was more comparable to the head volume, whereas the body is much larger. Nowadays, the most significant change in testing the PET counting rate performance is the change from the 19-cm-long phantom to a 70-cm-long phantom, while keeping the diameter (20 cm) of both phantoms unchanged. The 70-cm phantom is a better approximation to the activity distribution in whole-body studies, because the effects of out-of-field activity (OFA) are included in the measurements. The 19-cm, on the other hand, continues to be used to test the performance of scanners used primarily for brain imaging, particularly for scanners that are dedicated brain imaging instruments.

Most patient PET studies are not performed under conditions of low counting rate losses or negligible random rates. At higher activity levels, coincidence events are lost because of system dead time, whereas the rate of random coincidences rises. It is necessary to measure the counting rate performance, both dead-time losses and random events, as a function of activity to understand the scanner’s behavior for a wide range of scanning conditions.

Different studies have examined the axial and transverse spatial resolution within the field-of-view (FOV) measured with a point source. These studies, however, do not take into account the effect of varying activities on the resolution. The increase in count rate increases the probability that pulses from multiple coincidences will be
integrated together to cause spatial distortion and degrade the spatial resolution. The consequences of high count rate on spatial resolution are therefore investigated.

The aim of this study was to assess the count rate capabilities of the 3D dedicated PET scanner and determine the effects of the high count rate on the system’s spatial resolution. To quantify the changes in spatial resolution, the full width at half maximum (FWHM) of line source images and modulation transfer functions (MTF) were computed from images of a decaying line source in the 70 cm cylindrical phantom. The practicality of obtaining both count rate and spatial resolution data in a single acquisition was also determined.
CHAPTER 2
BACKGROUND

Count Loss and Dead Time

An incident 511-keV annihilation photon that is absorbed in the scintillator generates a pulse of light that is converted into an electronic signal and amplified by the photomultiplier tubes (PMTs). The outputs from the photomultipliers are used to localize the incident photon and as a measure of the energy to reject photons that have scattered before reaching the detector.

The overall count rate performance of the detectors is dependent on a number of factors such as pulse pile-up and system dead time or pulse resolving time $\tau$. Pile-up within the crystal occurs when two photons from different annihilations arrive so closely spaced in time that they cannot be distinguished as two separate photons. The light output of such an event is the sum of the two photon energies that will, in general, exceed the upper energy level discriminator (ULD), and therefore be discarded. Both photons are subsequently lost. This type of signal pile up is called post-pulse pile-up because a pulse is received while another is being integrated. A second source of count rate loss, known as pre-pulse pile-up, occurs when the positional and energy determination of a photon is still in process when a second photon arrives. Since the detection system is dead when the second photon arrives, the count will be lost. Such a process has the characteristics of a saturating or paralyzing system. A paralyzable system is one for which each event introduces a dead time whether or not that event actually was counted. Thus an event occurring during the dead time of a preceding event would not be counted but still would
introduce its own dead time during which subsequent events could not be recorded. To reduce pulse pile-up at high count rates, scintillators with short decay times are essential. The decay constant for the Gadolinium Oxyorthosilicate (GSO) scintillation crystal considered in this study is 56 nanoseconds. Scintillation light of GSO is emitted faster than Sodium Iodide (NaI(Tl)). NaI(Tl) has long been used as a scintillation crystal in most nuclear medicine imaging systems. NaI has a large decay constant of 230 nanoseconds. To further reduce pile-up and improve the high-count rate performance of the system, the amplifier shapes the pulse thereby reducing it to 100 nanoseconds [Fig. 1].

**FIG. 1.** Pulse before and after shaping

Because of dead time losses, the observed counting rate $R_o$ is less than the true counting rate $R_t$, where the latter is the counting rate that would be recorded if there was no dead time and is proportional to the source activity. At low activity, an increase in count rate is proportional to the increase in activity. However, at high activity levels, the proportionality is lost, as the system is unable to handle the increasing count rate. At sufficiently high activity levels, the count rate actually decreases with increasing activity.
as the system becomes paralyzed. The relationship between the observed and true
counting rates and the paralyzable dead time is given as follows:

\[ R_0 = R_1 e^{-R_1 \tau} \]  \hspace{1cm} (1)

Notice that the observed counting rate rises to a maximum value given by

\[ R_0^{\text{max}} = \frac{1}{2.718 \tau} \]  \hspace{1cm} (2)

Then the observed counting rate actually decreases with a further increase in true
counting rate. This is because additional events serve only to extend the already long
dead time intervals without contributing to additional events in the observed counting
rate.

The equation for the degradation of count rate due to pre-pulse dead time can be
extended to coincidence counts by noting that both annihilation photons must be
detected. If only one of the paired photons is detected, a coincidence event will not be
registered but instead will be lost. Therefore, the coincidence count rate, \( R_c \), can be
expressed as

\[ R_c = f R_1 e^{-R_1 \tau} e^{-R_1 \tau} = f R_1 e^{-2R_1 \tau} \]  \hspace{1cm} (3)

where \( f \) is the fraction of the true single event rate, \( R_1 \), which will create coincidence
events. Note that for the coincidence case, \( R_1 \) refers to the rate at which photons interact
with the scintillation crystal, and is assumed equal at both detectors.

Similarly, the equation for the degradation of coincidence count rate due to post-
pulse dead time is given as

\[ R_c = f R_1 e^{-2R_1 \tau} e^{-2R_1 \tau_n} = f R_1 e^{-2R_1 (\tau + \tau_n)} \]  \hspace{1cm} (4)

where \( \tau_n \) is the integration time and the sum \( (\tau + \tau_n) \) is the effective dead time constant.
The Effects of High Count Rate on Spatial Resolution

As pulses arrive after the integration process has begun, the extraneous light may be integrated with the original pulse. The extra signal contains additional spatial information that, when integrated with the spatial information of the original signal, may misplace the gamma ray’s origin in the patient. This misplacement will create spatial distortion and worsen spatial resolution. The inaccurate positional information is reduced by the post-pulse energy discrimination levels. An alternative to energy thresholds employed by the detection system, local centroid algorithm, and hence local centroid position, is used. The local centroid algorithm ensures that the PMTs that may receive light from a pile up event are excluded from the position calculation.

Random and Scatter Coincidence Events

In addition to detection of the true coincidence events, PET imaging with coincidence detection can result in two other undesirable types of events; scatter and random events. Scattered coincidences occur when one or both of the gamma rays undergo a Compton scatter interaction inside the body. This process changes their direction and reduces the energy of the photon. The change in direction results in misidentification of the gamma-ray origin as shown in Figure 2B. The fraction of gamma rays which get scattered depends on the scattering media and path length through the body. Therefore, the contribution from scattered events is more evident in abdominal imaging than in brain imaging. Although the gamma rays that are scattered have their energy reduced below 511 keV, the energy resolution of most PET systems is insufficient to use this as an effective means of scatter rejection. Thus, many of the scattered events are accepted and subsequently lead to falsely positioned data despite the scatter correction techniques.
Random, or accidental, coincidences arise when two photons from different annihilations are detected within the coincidence window and recorded as a coincidence. This situation may arise either when the partner photons are scattered out of the FOV [Fig. 2C], or when the two uncorrelated photons simply arrive more closely in time than the true coincidence. The random coincidence rate increases with the singles rate on the detectors because the probability that two uncorrelated photons will arrive within coincidence window increases. The random coincidence rate is proportional to the square of the activity and is, therefore, a particular problem for high count rate studies. The randoms rate \( R \) is strictly related to the singles counting rate \( S_1 \) and \( S_2 \) of each detector and to the coincidence time window width \( \tau_c \) by the following relation \(^1\!^9\nabla\):

\[
R = 2 \tau_c S_1 S_2
\]

Therefore, a quadratic increase in random events will be observed by increasing the radioactivity, whereas a decrease will result from reducing \( \tau_c \). Random coincidences contribute to the background in the image, which can lead to a loss of image contrast.
Efforts have been made to increment the net trues counting rate without incrementing random coincidences, by implementing faster systems with narrower coincidence time windows. The coincidence time window for the GSO crystal is 8 nanoseconds.

**Noise Equivalent Count Rate**

The relative noise in an acquisition is the standard deviation $\sigma$ of the total counts divided by the mean value $N$ of the total counts. Because counts in a nuclear medicine acquisition follow a Poisson distribution, the relative noise is simply $1/\sqrt{N}$. When random and scatter coincidence events are removed to determine the true count rate, the mean counts are reduced but the standard deviation remains the same assuming no change in randoms or scatters. The distribution is no longer Poisson and the relative noise is now $\sqrt{N}/(N-A)$, where $A$ is a constant representing the random and scatter coincidence events that were removed from the total counts.

The noise equivalent count (NEC) rate is a useful parameter indicating the noise component of PET data. It is a useful predictor of the PET scanner performance because it combines the effects of signal and noise in the system in a single metric. NEC rate provides a quantitative framework in which to analyze design options that may increase true, scattered, and random coincidences, by varying degrees, to assess their impact on imaging performance.

The NEC rate represents the ratio between net trues and prompts. The NEC rate follows Poisson statistics and has a relative noise equal to the relative noise of the true events after the random and scatter coincidence events have been removed. Setting the relative noise equal to $1/\sqrt{\text{NEC}}$, then the NEC rate is given by the following equation:

$$\text{NEC} = \frac{(N - A)^2}{N}$$ (6)
It is worth mentioning that the best imaging condition is reached when acquisitions are performed by placing an activity in the FOV corresponding to the peak NEC rate.
CHAPTER 3
MATERIALS AND METHODS

Camera Description

The camera used was a Philips Allegro Positron Emission Tomography (PET) system. The system is equipped with 420 Photo-Multiplier Tubes (PMTs) and 28 modules. Full-ring detection geometry is defined by placing the modules side by side, coupled to the PMTs closely packed in a hexagonal array. Each module is populated with an array of 638 Gadolinium Oxyorthosilicate (GSO) crystals, totaling 17,864 crystals. Each crystal has dimensions of 4x6x20 mm. The modules are arranged around the patient port with the GSO crystals facing the patient. The 420 PMTs are mounted on the opposite side of the 28 modules which collect light from the scintillation crystals [Fig. 3]. By surrounding the patient with 28 position-sensitive segments, data can be acquired simultaneously along any parallel line and at any right angle.

FIG. 3. Module Assembly
The axial field-of-view (FOV) of the camera is 18 cm and the transverse FOV is 57.6 cm. The system operates in 3D mode, and therefore no physical collimators are involved. The scanner houses a $^{137}$Cs source, 662 keV, for generating transmission images during patient scanning. The transmission data is used to correct the emission images for photon attenuation in the patient’s body. Transmission scans are not used in this investigation because, according to the National Electrical Manufacturers Association\textsuperscript{16} (NEMA), attenuation corrections should not be employed when measuring the count rate capability of the scanner. NEMA is further discussed in the following sections.

**Phantom Description**

The phantom considered here is described in section 4 of the National Electrical Manufacturers Association (NEMA) NU 2-2001 document of performance standards for PET scanners.\textsuperscript{16} A list of the NEMA protocol is provided in Appendix A. The test phantom [Fig. 4(A), and 4(B)] is a solid circular cylinder, as shown below,

FIG. 4. Phantom utilized in the experiment; (A) Phantom and line source, (B) Cross section of the cylindrical phantom showing line source inserted 4.5 cm off center.
composed of polyethylene with an outside diameter of 20 cm and with an overall length of 70 cm. A 6.4 mm hole is drilled parallel to the central axis of the cylinder, at a radial distance of 4.5 cm. The cylinder consists of 4 segments that are assembled together during testing. The assembly of the completed phantom was checked to insure a tight fit between adjacent segments, as even very small gaps will allow narrow axial regions of scatter-free and attenuation-free radiation. The test phantom line source insert is a clear polyethylene coated plastic tube that is 80 cm in length, with an inside diameter of 3.2 mm and an outside diameter of 4.8 mm. The central 70 cm of this tube was filled with an $^{18}$F-FDG initial activity of 19.26 millicuries (mCi) at 3:30 PM and threaded through the 6.4 mm hole in the test phantom. Notice that the phantom in this experiment is longer than the axial FOV of the scanner, which allows studying of the effects of out-of-field activity on the count rate. For consistency, the phantom was rotated such that the line source is at the lowest position (i.e., nearest to the patient’s table), because the measured result will depend on the relative orientation of the line source and the table.

**Allegro Data Acquisition and Processing**

The phantom was centered in the 18 cm field of view of the scanner and the dynamic NEMA-Countloss protocol was used to acquire data at different activities. Each acquisition lasted for 20 minutes. A delay time of 20 minutes was used between acquisitions. The first acquisition began at 4:38 PM with a calculated initial activity of 12.536 mCi. A total of 21 tomographic acquisitions were taken every 40 minutes. Table I, in the results chapter, lists the initial conditions of the image acquisitions, including the times of the acquisitions, the initial activities at the beginning of each acquisition, and the time into the experiment. A complete description on how to acquire data using the Allegro PET scanner is given in the following sections.
**Allegro Data Acquisition Parameters**

The Allegro workstation consists of two monitors; one for acquisition, and another for processing. File Management was chosen from the acquisition monitor. Invoked within the File Management is the Acquisition menu that is used to setup the parameters for the acquisition. By choosing Set Up Acquisition, one can enter all information pertinent to that acquisition, such as study date, phantom weight, etc. The information required here are easy to understand and require no previous experience. Notice that the directory where the output raw data will reside does not need to be specified. After the acquisition is finished, the raw data will be transferred to the patient directory automatically and will be easily identified. However, the filename should be entered to be able to identify the correct file. For further clarification on the acquisition parameters and any other software setup information, one can refer to the Allegro user’s manual. The dynamic DefaultNEMA_Countloss is selected as the acquisition protocol. Typically, acquisitions are performed using a preset protocol. In this experiment, however, it was necessary to change some of the acquisition parameters, which is done by selecting to edit the protocol. The editing process consists of specifying the acquisition time and adding more frames (or acquisitions) at the end until a small activity is left in the line source. The number of frames added depends on the starting activity and the acquisition time chosen. The durations of all the frames could be added to get a good prediction of what the last frame should be. After acquiring, a file containing the raw data is generated and saved in the patients’ directory. This file is further processed as shown in the following section.
Allegro Data Processing Parameters

The raw data file, the output at the end of the last acquisition, is then processed on the processing monitor. To do so, the NEMA_Countloss reconstruction protocol is selected by highlighting the file, and choosing Petview, Reconstruct Sinogram, and Research Protocol, respectively. The reconstruction protocol is then edited to generate the Fourier rebinned\textsuperscript{18,19} data to be use in the data analysis programs. The parameters of NEMA_Countloss reconstruction protocol are already set correctly as specified by the NEMA standards for testing the count rate performance of positron emission tomographs. Examples of these parameters are the background subtraction, attenuation correction, and decay correction, which are all turned off during the reconstruction. However, the user needs to save the Fourier rebinned (FORE) sinograms using the FT Output Sinogram option, found in the advanced parameters menu. A filename can also be specified for the rebinned data. Notice that in the advanced parameters, there is an option provided by the manufacturer to turn the Single Slice Rebinning\textsuperscript{18} (SSRB) on. The SSRB, however, was not used because it is believed that this rebinning algorithm is not functioning at all. Despite consultation with the software experts at Philips, this matter has not been resolved. The manufacturer, however, suggested using the FORE for this experiment.

File Structure

The reconstruction program operates on the raw data to produce an interpolated scan file. The structure of this interpolated file is as follows. It has a main header that is 512 bytes long. As a confirmation, the headers display acquisition parameters and patient information. The main header contains information such as the date of the experiment or patient scan, the patient ID, isotope used, etc. The main header is followed by a directory record, which is 512 bytes long. Every directory record is followed by 31 slices. If the
file contains more than 31 slices, a new directory record is created after slice number 31. Sinograms and directory records are added to the file as needed until all data are stored. Each slice or sinogram within the file has a subheader. The size of the subheader is also 512 bytes followed by data. The data in the sinogram are always written as signed short (2 bytes) variables. The sinograms are treated as 256 by 192 arrays.

The overall structure of the Fourier rebinned file is the same as the interpolated file. However, the interpolated file contains 45 tilted slices at each tilt angle. There are 7 tilt angles for each sinogram. Thus, there are 7 tilted slices at each position along the axis parallel to the axial field-of-view, including the zero tilted sinogram or direct sinogram. These oblique sinograms are collapsed into a single sinogram for each respective slice using the FORE algorithm to produce only direct sinograms suitable for NEMA processing. Therefore, reducing the total number of slices from 315 to 45 slices per acquisition. The Fourier rebinned file and the interpolated file are both output by the reconstruction protocol.

To recap, in the interpolated file each event is binned into 4-dimensional projection coordinates; Transverse distance r, projection angle $\phi$, tilt or off-plane angle $\theta$, and axial distance $y'$. FORE is then employed to convert these 4-dimensional sinograms into 2-dimensional sinograms by collapsing the oblique sinograms into a single sinogram for each respective slice while conserving the number of counts in the sinogram. The resulting sinogram is a 256 by 192 pixel matrix with a signed short format. The pixel’s linear size is 2.25 mm. No corrections for dead time, attenuation, or random events are employed during the acquisitions. A total of 45 sinograms per acquisition are obtained in order to span the length of the line source in the 18 cm field of view.
In addition, the Fourier rebinned file is then copied to a specified directory to be burned on a CD. Appendix B provides more information on how to copy and burn files to a CD.

All pixels in a sinogram are multiplied by a scaling factor to correct the number of counts in each pixel. This is done when running the MATLAB codes, discussed in the following section. The scaling factors are obtained using a shell script provided by the software engineer at Philips Medical Systems [appendix C], which is a small program that can be run in a Unix environment. The output from this program is three columns. The first is frame number, running from 1 to 21. The second is a slice index, typically running from 1 to 45. The third value is the sinogram slice scale factor. To run this program in Unix, one should refer to the commands shown in appendix C.

**NEMA Data Processing and Analysis Using MATLAB**

**Count Rates**

MATLAB codes, written by the author of this paper, were employed to perform the NEMA data analysis for processing the sinograms data obtained after the experiment. These codes are given in appendices D through I and will be briefly discussed here. Notice that comments are embedded within each code to make them easy to understand and user friendly. These MATLAB codes need to be run successively. Because the output from the first program [appendix F] will become the input to the second program [appendix G] and the output from the second program will become the input to the third program [appendix H], and so on. Appendix D shows the code for extracting the sinograms from the raw data file. The extracted sinograms of each acquisition are output.

† MATLAB: The language of technical computing by MathWorks, Inc. 2002
to different acquisition files numbered 1 through 21. The main header of the original file, the subheader of each sinogram, and the directory records are all not included in the output acquisition files. Therefore, the output data are only the sinograms data needed for processing. This step of reorganizing the data such that no header information is embedded within the actual sinogram data is significant for it reduces the amount of code necessary in processing the sinograms. Appendix E has the scaling factors code. This is a simple piece of code written to organize the scaling factors in a format suitable for processing when used by the programs to follow. Again, the order in which these programs are run is important and follows the order shown in the appendices.

Section 4 of the NEMA standards publication NU 2-2001 for performance measurements of positron emission tomographs summarizes the steps used to determine scatter fraction, count losses, and randoms measurements. The MATLAB codes in appendices F and G were both used for further processing of the data. In these codes, pixels that were 12 cm or further away from the center of the phantom were set to zero counts for each sinogram. For each projection angle within a sinogram, the location of the line source’s center was determined as the pixel where the highest count value occurred. The count data within that projection angle were then shifted in the transverse direction so that the center of the line source aligned with the center of the sinogram in the transverse direction [Fig. 5]. After the shift, each sinogram was compressed into a one-dimensional profile, or sum projection, by summing the pixel values along the angular direction.
In slight deviation from NEMA, these profiles were then summed over all sinograms for the particular acquisition to create a final sum projection to describe each acquisition. It was from these final sum projections, rather than the sum projections of the individual slices, that the count rates were determined. From these count rates, the activity where the peak rate occurs was found.

Figure 6 illustrates how the random and scatter coincidence events were determined for each acquisition. The counts at the pixels 2 cm to either side of the center of the final sum projection were averaged and then multiplied by the number of pixels equal to a distance of 4 cm. This product was added to counts outside the 4 cm strip to create the...
random plus scatter counts for acquisition j, \( C_{r+s,j} \). To separate the scatter and random components, the final acquisition j’, where the random coincidence rate and the loss in count rate due to dead time are both presumed less than 1%, was considered. With the dead time equation and equations for coincidence count rates\(^7,11\), it was verified that the dead time losses and the random count rates in the final acquisition would both be less than one percent. Letting the random coincidence counts be negligible, the scatter and random counts, \( C_{r+s,j} \) were divided by the total counts of acquisition j’, \( C_{TOT,j} \) to calculate the scatter fraction SF:

\[
SF = \frac{C_{r+s,j}}{C_{TOT,j}}
\]  

(7)

This scatter fraction was constant for all acquisitions because the geometry and attenuation media remained constant. Variations in design cause PET scanners to have different sensitivities to scattered radiation. The scatter fraction is measured at a sufficiently low counting rate that random coincidences, dead-time effects, and pileup are negligible. The total event rate, \( R_{TOT,j} \) for each acquisition is the total counts for that acquisition divided by the acquisition time \( T_{acq,j} \):

\[
R_{TOT,j} = \frac{C_{TOT,j}}{T_{acq,j}}
\]  

(8)

The true event rate, free of scatter or random coincidences, is

\[
R_{t,j} = \frac{C_{TOT,j} - C_{r+s,j}}{T_{acq,j}}
\]  

(9)

The scatter and random count rates can be found by using the following equations:

\[
R_{s,j} = \frac{SF}{1 - SF} R_{t,j}
\]  

(10)
\[ R_{t,j} = R_{TOT,j} - \frac{R_{t,j}}{1 - SF} \]  \hspace{1cm} (11)

The noise equivalent count rate was determined by the following equation:

\[ R_{NEC,j} = \frac{R_{t,j}^2}{R_{TOT,j}} \]  \hspace{1cm} (12)

To determine if the random coincidence count rate increases with the square of the activity, a function was fitted to the plot of computed random count rates as a function of activity [appendix H]. The fitted function was of the following form:

\[ F(x; a) = ax^2 \]  \hspace{1cm} (13)

where \( x \) is the activity in millicuries (mCi), and \([a]\) is a constant.

**Spatial Resolution**

The final sum projection of each acquisition was fitted to a Gaussian function plus a constant, equation 14, using the nonlinear least squares method in MATLAB [appendix H]. To further verify the fitted curves, the final sum projection of each acquisition was plotted on the software package Kaleida Graph 3.5\(^\dagger\) and a function was fit to the plot. An online curve-fitting tool\(^\ddagger\) was also used for the same purpose. The fitted function that was always considered was a gaussian plus a constant. The parameters of the fitted functions were all the same within three significant figures. The Kaleida Graph parameters, however, were slightly different. The FWHMs, calculated based on the fitted parameter of the standard deviation, are, therefore, only rough calculations due to errors that are likely to be associated with the fitting process.

\(^\dagger\) A trial version of this software package was used.

\(^\ddagger\) http://zunzun.com: An interactive 2-D and 3-D data-modeling tool.
The spatial resolution was then determined by calculating the full width at half maximum (FWHM) for each acquisition using equation 15. The standard deviation $\sigma$ in equation 15 is obtained from the fitted functions. The FWHM was adjusted for the finite diameter of the line source (3.2 mm) using equation 16. This equation was based on the practice of squaring components of resolution, adding the squared components together, and then taking the square root of this sum to determine the final resolution.

$$F(x; a, b, x_0, \sigma) = a + b e^{-0.5(x-x_0)^2/\sigma^2}$$  \hspace{1cm} (14)

$$\text{FWHM} = 2.35 \sigma$$  \hspace{1cm} (15)

$$\text{FWHM}_a(\text{mm}) = [(\text{FWHM}(\text{mm}))^2 - (3.2)^2]^{1/2}$$  \hspace{1cm} (16)

To further explore the spatial resolution of an imaging system, the modulation transfer function (MTF) is preferred. By using MATLAB [appendix I], an MTF was calculated for each acquisition by computing the discrete Fourier transform of the final summed projection of each acquisition. The MTFs were each normalized to a value of unity at the zero frequency. This analysis on the spatial resolution is confined to only the resolution in the transverse direction because of the type of geometry and analysis followed.
CHAPTER 4
RESULTS

Table 1 lists the initial conditions of the image acquisitions, including the times of the acquisitions, the initial activities at the beginning of each acquisition, and the time into the experiment.

<table>
<thead>
<tr>
<th>Acquisition</th>
<th>Start time</th>
<th>End time</th>
<th>Time into experiment (min)</th>
<th>Initial activity (mCi)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>4:38 PM</td>
<td>4:58 PM</td>
<td>0</td>
<td>12.536</td>
</tr>
<tr>
<td>2</td>
<td>5:18 PM</td>
<td>5:38 PM</td>
<td>40</td>
<td>9.738</td>
</tr>
<tr>
<td>3</td>
<td>5:58 PM</td>
<td>6:18 PM</td>
<td>80</td>
<td>7.564</td>
</tr>
<tr>
<td>4</td>
<td>6:38 PM</td>
<td>6:58 PM</td>
<td>120</td>
<td>5.876</td>
</tr>
<tr>
<td>5</td>
<td>7:18 PM</td>
<td>7:38 PM</td>
<td>160</td>
<td>4.564</td>
</tr>
<tr>
<td>6</td>
<td>7:58 PM</td>
<td>8:18 PM</td>
<td>200</td>
<td>3.545</td>
</tr>
<tr>
<td>7</td>
<td>8:38 PM</td>
<td>8:58 PM</td>
<td>240</td>
<td>2.754</td>
</tr>
<tr>
<td>8</td>
<td>9:18 PM</td>
<td>9:38 PM</td>
<td>280</td>
<td>2.139</td>
</tr>
<tr>
<td>9</td>
<td>9:58 PM</td>
<td>10:18 PM</td>
<td>320</td>
<td>1.662</td>
</tr>
<tr>
<td>10</td>
<td>10:38 PM</td>
<td>10:58 PM</td>
<td>360</td>
<td>1.291</td>
</tr>
<tr>
<td>11</td>
<td>11:18 PM</td>
<td>11:38 PM</td>
<td>400</td>
<td>1.003</td>
</tr>
<tr>
<td>12</td>
<td>11:58 PM</td>
<td>12:18 AM</td>
<td>440</td>
<td>0.779</td>
</tr>
<tr>
<td>13</td>
<td>12:38 AM</td>
<td>12:58 AM</td>
<td>480</td>
<td>0.605</td>
</tr>
<tr>
<td>14</td>
<td>1:18 AM</td>
<td>1:38 AM</td>
<td>520</td>
<td>0.470</td>
</tr>
<tr>
<td>15</td>
<td>1:58 AM</td>
<td>2:18 AM</td>
<td>560</td>
<td>0.365</td>
</tr>
<tr>
<td>16</td>
<td>2:38 AM</td>
<td>2:58 AM</td>
<td>600</td>
<td>0.284</td>
</tr>
<tr>
<td>17</td>
<td>3:18 AM</td>
<td>3:38 AM</td>
<td>640</td>
<td>0.220</td>
</tr>
<tr>
<td>18</td>
<td>3:58 AM</td>
<td>4:18 AM</td>
<td>680</td>
<td>0.171</td>
</tr>
<tr>
<td>19</td>
<td>4:38 AM</td>
<td>4:58 AM</td>
<td>720</td>
<td>0.133</td>
</tr>
<tr>
<td>20</td>
<td>5:18 AM</td>
<td>5:38 AM</td>
<td>760</td>
<td>0.103</td>
</tr>
<tr>
<td>21</td>
<td>5:58 AM</td>
<td>6:18 AM</td>
<td>800</td>
<td>0.080</td>
</tr>
</tbody>
</table>
Count Rate

The final sum projections of acquisitions 1, 5, 10, 15, and 21 are shown in Figure 7.

![Count Rate Graph]

**FIG. 7.** Final sum projections of selected acquisitions

Figure 7 illustrates the changes in count rate, width of the peak, and background level due to random and scattered events. The background decreases over time indicating that the random and scatter coincidences together are decreasing. The final sum projections of all acquisitions are shown in Figures 8 through 28 [Appendix J]. Notice that the y-axis of Figures 8 through 28 is changed to reflect the changes in the count rates for each separate acquisition.

The total paired events, and total, true, scatter, random, and NEC count rates are all shown in Table 2. The scatter fraction was calculated to be 0.403. This scatter fraction was computed from the last acquisition.
FIGURE 29. Illustration of the counting rate performance of the PET scanner.

The total, true, random, and scatter count rates are plotted in Figure 29 as functions of time into the experiment, and in Figure 30 as functions of activity. The true count rate is plotted as a function of time into experiment in Figure 31, and as a function of activity in Figure 32. As can be observed in Figures 31 and 32, the true count rate reaches a maximum of 83.86 kcps at an activity of 9.738 mCi. The noise equivalent count rate is plotted as a function of time into experiment in Figure 33 and as a function of activity in Figure 34. Notice that the noise equivalent count rate also reaches a maximum and drops again as activity increases. The maximum NEC rate is 31.04 kcps at 5.876 mCi. As shown in the figures, the total, true, and scatter count rates will all rise with activity.
FIG. 29. Count rates versus time into experiment

FIG. 30. Count rates versus activity
FIG. 31. True count rate versus time into experiment

FIG. 32. True count rate versus activity
FIG. 33. Noise Equivalent Count rate versus time into experiment

FIG. 34. Noise Equivalent Count rate versus activity
and tend to reach a maximum value before they start to decrease. The total peak count rate is not shown here because higher activities are needed to reach the maximum. The shape of these count rate curves is consistent with the behavior of the radiation detector that experiences paralyzable dead time and pulse pile up. The count rate increases, starting at low activities, until a maximum count rate is reached and then starts to decrease as the activity continues to grow. The random coincidence count rate, on the other hand, starts very small at low activities and continues to increase with increasing activity. As expected, the continued increase in randoms reflects the fact that random coincidence rate is proportional to the square of the activity. A fact that is also illustrated by the fitted function [see Equation 13]. Figure 35 below shows the fitted randoms as a function of activity. The constant [a] of Equation 13 was found to have a value of 1028 cps and the fit had an R-squared value of 0.976.

![FIG. 35. Randoms count rate and its fitted equation versus activity](image-url)
Spatial Resolution

The parameters of the fitted Gaussian equation, Equation 14, of the final sum projection are given in Table 3 for each acquisition.

<table>
<thead>
<tr>
<th>Acquisition</th>
<th>a [cps]</th>
<th>b [cps]</th>
<th>X₀ [cm]</th>
<th>σ [cm]</th>
<th>R-squared</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>770.44</td>
<td>22650.41</td>
<td>-0.0013</td>
<td>0.35401</td>
<td>0.888</td>
</tr>
<tr>
<td>2</td>
<td>615.53</td>
<td>23107.80</td>
<td>-0.0025</td>
<td>0.34175</td>
<td>0.923</td>
</tr>
<tr>
<td>3</td>
<td>481.14</td>
<td>22195.66</td>
<td>-0.0022</td>
<td>0.33342</td>
<td>0.944</td>
</tr>
<tr>
<td>4</td>
<td>371.08</td>
<td>20343.87</td>
<td>-0.0031</td>
<td>0.32724</td>
<td>0.957</td>
</tr>
<tr>
<td>5</td>
<td>284.04</td>
<td>17966.86</td>
<td>-0.0029</td>
<td>0.32270</td>
<td>0.965</td>
</tr>
<tr>
<td>6</td>
<td>217.22</td>
<td>15439.59</td>
<td>-0.0023</td>
<td>0.31935</td>
<td>0.971</td>
</tr>
<tr>
<td>7</td>
<td>165.78</td>
<td>12959.74</td>
<td>-0.0029</td>
<td>0.31679</td>
<td>0.975</td>
</tr>
<tr>
<td>8</td>
<td>126.57</td>
<td>10691.55</td>
<td>-0.0031</td>
<td>0.31486</td>
<td>0.977</td>
</tr>
<tr>
<td>9</td>
<td>97.07</td>
<td>8712.33</td>
<td>-0.0021</td>
<td>0.31341</td>
<td>0.979</td>
</tr>
<tr>
<td>10</td>
<td>74.19</td>
<td>7001.64</td>
<td>-0.0027</td>
<td>0.31233</td>
<td>0.980</td>
</tr>
<tr>
<td>11</td>
<td>57.03</td>
<td>5593.31</td>
<td>-0.0028</td>
<td>0.31169</td>
<td>0.981</td>
</tr>
<tr>
<td>12</td>
<td>43.89</td>
<td>4433.85</td>
<td>-0.0023</td>
<td>0.31111</td>
<td>0.982</td>
</tr>
<tr>
<td>13</td>
<td>33.87</td>
<td>3506.68</td>
<td>-0.0028</td>
<td>0.31039</td>
<td>0.983</td>
</tr>
<tr>
<td>14</td>
<td>26.17</td>
<td>2763.25</td>
<td>-0.0021</td>
<td>0.31002</td>
<td>0.983</td>
</tr>
<tr>
<td>15</td>
<td>20.22</td>
<td>2163.63</td>
<td>-0.0026</td>
<td>0.30961</td>
<td>0.983</td>
</tr>
<tr>
<td>16</td>
<td>15.75</td>
<td>1700.44</td>
<td>-0.0034</td>
<td>0.30964</td>
<td>0.984</td>
</tr>
<tr>
<td>17</td>
<td>12.24</td>
<td>1331.75</td>
<td>-0.0018</td>
<td>0.30946</td>
<td>0.984</td>
</tr>
<tr>
<td>18</td>
<td>9.51</td>
<td>1039.44</td>
<td>-0.0018</td>
<td>0.30927</td>
<td>0.984</td>
</tr>
<tr>
<td>19</td>
<td>7.44</td>
<td>813.13</td>
<td>-0.0034</td>
<td>0.30927</td>
<td>0.984</td>
</tr>
<tr>
<td>20</td>
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<td>628.40</td>
<td>-0.0031</td>
<td>0.30926</td>
<td>0.984</td>
</tr>
<tr>
<td>21</td>
<td>4.54</td>
<td>494.66</td>
<td>-0.0022</td>
<td>0.30935</td>
<td>0.984</td>
</tr>
</tbody>
</table>

Figures 36 through 56 [Appendix K] show the fitted final sum projections. Figures 36 and 56 are shown on the next page to illustrate the differences in the fit between the first and last acquisitions. This approach of fitting the final sum projections to a gaussian function plus a constant is not very accurate. The fitting parameters would differ when a better fit is introduced. Therefore, the FWHM would change as well. Due to this inaccurate curve fitting, the technique employed here remains subject to error. However, it still provides a rough estimation of the resolution to at least get an overall picture of the imaging system behavior. The calculated FWHMs are given in Table 4. The adjusted
FIG. 36. Final sum projection of acquisition 1 and its fitted gaussian curve

FIG. 56. Final sum projection of acquisition 21 and its fitted gaussian curve
TABLE 4. FWHM calculated from the fitted final sum projections

<table>
<thead>
<tr>
<th>Acquisition</th>
<th>measured FWHM [mm]</th>
<th>adjusted FWHM [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>8.32</td>
<td>7.68</td>
</tr>
<tr>
<td>2</td>
<td>8.03</td>
<td>7.37</td>
</tr>
<tr>
<td>3</td>
<td>7.84</td>
<td>7.15</td>
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<tr>
<td>4</td>
<td>7.69</td>
<td>6.99</td>
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<tr>
<td>5</td>
<td>7.58</td>
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<td>6</td>
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<td>7</td>
<td>7.44</td>
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<td>8</td>
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<td>9</td>
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<td>6.54</td>
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<tr>
<td>17</td>
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</tr>
<tr>
<td>21</td>
<td>7.27</td>
<td>6.53</td>
</tr>
</tbody>
</table>

FWHMs are plotted as functions of time into experiment and activity as shown in Figures 57 and 58, respectively. The degradation in the spatial resolution can be seen as the FWHM increases from 6.53 mm to 7.68 mm as the activity increases. After the 16th acquisition, no further improvement is seen in the spatial resolution. The FWHM has a constant value of 6.53 mm as activity continues to decay going from the 17th acquisition to the 21st acquisition. The spatial resolution of the PET scanner at 10 cm from the center in the transverse direction is 5.9 mm, as measured by a point source. This value was provided by Philips-ADAC Medical Systems. The different value obtained for the resolution, therefore, could have been a result of the different approach used to calculate the resolution of the camera and the finite line source diameter involved in the
FIG. 57. Adjusted FWHM as a function of time into experiment

FIG. 58. Adjusted FWHM as a function of activity

measurement. The point to keep in mind here is that the spatial resolution improves as the activity decreases. Therefore, the administration of entailed high activity levels to the
patients will not only increase their radiation absorbed dose, but also degrades the spatial resolution.

Assessment of the spatial resolution is also obtained from the modulation transfer functions. Figure 59 shows the modulation transfer functions of acquisitions 1, 5, 10, and 21. For the first acquisition, at an activity equal to 12.536 mCi, the gain at a spatial frequency of 0.1 cm\(^{-1}\) is only 24.3%. This gain increases in the fifth acquisition, which has an activity of 4.564 mCi, to 42.6% at the same spatial frequency. Figures 60 through 80, in Appendix L, are the modulation transfer functions for each acquisition. Acquisitions 10, 15, and 21 also show the increase in gain to 52.8%, 56%, and 56.5%, respectively, at 0.1 cm\(^{-1}\) spatial frequency. The MTF Figures also illustrate the same trend of increasing gain across the range of frequencies as activity decays over time. For example, at a frequency of 0.5 cm\(^{-1}\), the gain is only 15% at the first acquisition, but...
25.8% at the fifth, and 32.3%, 34.3%, and 34.6% for the tenth, fifteenth, and twenty-first acquisitions, respectively. No significant increase in the gain was noticed going from the seventeenth acquisition to the twenty-first. The gain for the latter acquisitions was almost constant and had a value of about 34.6%. This also explains the constant values obtained for the spatial resolution at the same last five acquisitions.

There was a ripple, or dip, present in each modulation transfer function. This rippling effect is believed to originate from the background counts that were included when computing the Fourier transforms. Therefore, exclusion of the background by subtracting the random and scatter events from the total events would eliminate or reduce the rippling significantly because the modulation transfer functions would now be created based on the true events only. Improvements in the gain, in this case, would also reflect the improvement in image contrast. The rippling in the MTF curves was first thought to be an artifact due to the truncation of counts located 12 cm away from the center of the image. However, the same effect was still seen in the MTFs, even without the truncation.
CHAPTER 5
DISCUSSION

Some recent studies have investigated the counting rate performance of state-of-the-art dedicated PET systems using different size phantoms.\textsuperscript{6, 7, 9} The most common of these phantoms were the 19-cm and the 70-cm phantoms. A longer phantom of 70-cm length was considered in order to study the effects of out-of-field activity on the count rates detected. The scatter fraction calculated from the last acquisition was 40.3\%. The scatter fraction is not representative of a realistic body size. However, it is useful for standard evaluations among scanners of different configurations. On the other hand, a concern with using the line source to measure the counting rate performance was whether or not it yields counting rates that are clinically relevant. A recent study\textsuperscript{7} shows that there is good correlation between the measured counting rates for the 70-cm line source and whole-body studies.

It is important in comparing performance between scanners to look both at the peak true and NEC counting rates and at the activity levels where these peak rates occur. A highly sensitive system may saturate at a relatively low activity level, but the counting rate at this level may be higher than that of a system with lower sensitivity at a higher activity level. The NEC rate peaked at 31.04 kcps, at an activity of 5.876 mCi, it then decreased with increasing activity. If the 70-cm phantom were considered to mimic the body of an adult, then an activity more than the 5.876 mCi should not be used because any increase in activity above that point would decrease the count rate due to pileup in the detector. According to Philips-ADAC Medical Systems, the peak NEC rate was 33
kcps at an activity of 0.25 µCi/mL or 5.5 mCi for the 22 liters phantom. The 22 liters is the volume of the 70-cm polyethylene phantom. In comparison, the manufacturer’s NEC rate was higher than the one calculated in this experiment because the value provided by the manufacturer was based solely on rough hand calculations. Moreover, the peak true count rate was 84 kcps at an activity of 0.4 µCi/mL or 8.8 mCi for the same phantom. This value was also based on rough hand calculations. These values were considered to be suggestive and not authoritative, as pointed out by the manufacturer.

The spatial resolution was degraded when activity increased, as can be seen from the FWHM plot as a function of activity. This should be taken into consideration when administering a radiopharmaceutical to the patient. The improvement in the FWHM is only 1.15 mm from the first to the last acquisition. The FWHM, where the peak NEC occurred, was 7mm. The true count rate at that resolution is 72.46 kcps. Better resolution can be obtained at lower activities, but the count rate would be noisier. A compromise, therefore, should be made between the resolution and the activity. The spatial resolution of the last five acquisitions remained the same as indicated by the FWHM value of 6.53 mm. This means that no further improvement in spatial resolution is obtained beyond the seventeenth acquisition. One could argue that the limit on the transverse spatial resolution of the scanner at 4.5 cm radial distance from the center of the transverse FOV is 6.53 mm. Further discussion on the modulation transfer functions that describes the spatial resolution of an imaging system was previously included in the results chapter of this paper. MTFs illustrated that the gain improves as activity decreases. Significant improvements in gain were shown up to the sixteenth acquisition, after which the gain remains unchanged. In the MTF plots, the curve extending from the zero frequency to the
region, where the little dip exists, is due to the scatter and random components. This portion of the plot corresponds to the background and appears to occur within the frequency range of zero to 0.125 cm\(^{-1}\). The other portion of the plot, extending from 0.18 cm\(^{-1}\) to the Nyquist frequency, 2.22 cm\(^{-1}\), is due to the true events.

Activity outside the field of view was shown to greatly affect the count rate performance, especially the scatter component, of the scanner as well as the spatial resolution. The random coincidence counts were mostly affected by the dead time of the radiation detector. The expected trend of the random events, as illustrated by the fitted equation in Figure 35, is a quadratic increase as the activity increases. However, the calculated random events continued to increase then their increase slows down as indicated by the intersection of the fitted and the measured random curves. This is due to the increased dead time of the radiation detector because of the more activity, and hence more counts, introduced. Each undetected event will only increase the dead time and therefore, some random counts will be lost.

Finally, the spatial resolution of the PET scanner degrades as more activity is administered to patients. In addition, more counts are lost at the same high activities. The optimal activity that should be administered to patients is the one at which the peak noise equivalent count rate occurs. Because very high activities will result in more counts being lost because of system dead time and counts pile up.
CHAPTER 6
CONCLUSIONS

The 3-dimensional scanning system had a high scatter fraction of 40%, compared to the scatter fraction obtained using a smaller phantom. This high value was expected because of the long phantom employed in the measurements and the 3-dimensional type system involved. The scattered events from outside the field of view contributed to this increase in the scatter fraction as opposed to smaller phantoms, typically 19-cm phantoms. The longer phantom provides a more clinically reasonable distribution of activity outside the scanner. This out-of-field activity can impact the counting rate performance as well as the spatial resolution. Future work involving the same measurements discussed in this investigation could be in the form of utilizing an even longer phantom, possibly 150 cm, which would mimic the length of an average person. This would more closely approximate a true whole-body clinical scan. The scatter fraction would also be expected to increase even more. However, the increased value would only mean that more scattered coincidences would be eliminated from the data reconstructed, and therefore, improve the overall quality of the clinical images. In addition, the NEMA protocol may be revised to include the new longer phantom, if proved to be better than the existing ones.

The peak noise equivalent count rate was determined to be 31.04kcps. This peak occurred at an activity of 5.876 mCi. The full width at half maximum decreased from 7.68 mm to 6.53 mm as activity decreased, indicating the improvement in spatial resolution as activity decays over time. The improvement in resolution was supported by
the modulated transfer functions. The study shows that both the count rate capabilities and the spatial resolution of the 3-dimensional dedicated positron emission tomograph will degrade at high levels of activity. This is important when determining the amount of radionuclide tracer that should be administered for a PET scan.

The amount of $^{18}$F radioactivity administered to patients undergoing a PET scan is usually about 6 mCi at the start of the acquisition and after enough time is allowed for uptake of the radionuclide in tissue. Notice that this value is too close to the peak NEC rate that occurs at 5.9 mCi. This amount will still introduce some amount of noise in the image, but it results in an ideal imaging condition. The NEC rate is the ratio between the net trues and prompts. Therefore, the best imaging condition is reached when acquisitions are performed using an activity corresponding to the peak NEC rate. In conclusion, to allow for more accurate imaging of patients, a compromise between the acceptable noise equivalent count rate and adequate spatial resolution is required.
NEMA Standards Publication NU 2-2001

*Performance Measurements of Positron Emission Tomographs*

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Section 4
SCATTER FRACTION, COUNT LOSSES, AND RANDOMS MEASUREMENT

4.1 GENERAL
The scattering of gamma rays emitted by the annihilation of positrons results in falsely located coincidence events. Variations in design and implementation cause positron emission tomographs to have different sensitivities to scattered radiation.

The measurements of count losses and random rates express the ability of a positron emission tomograph to measure highly radioactive sources with accuracy.


4.2 PURPOSE
The first purpose of this procedure is to measure the relative system sensitivity to scattered radiation. Scatter is expressed by the scatter fraction, SF, for the entire tomograph.

The second purpose of this procedure is to measure the effects of system dead-time and the generation of random events at several levels of source activity.

The true event rate is the total coincident event rate minus the scattered event rate and minus the randoms event rate. The method of randoms estimation includes misplaced events as part of the randoms measurement. Furthermore, shifts in the energy peak may influence the results for the randoms, since it is assumed that the scatter fraction is constant as a function of countrate.

4.3 METHOD
The test phantom is a solid right circular cylinder composed of polyethylene with a specific gravity of 0.96 ± 0.01, with an outside diameter of 203 ± 3 mm (8"), and with an overall length of 700 ± 5 mm. A 6.4 ± 0.2 mm (1/4") hole is drilled parallel to the central axis of the cylinder, at a radial distance of 45 ± 1 mm. For ease of fabrication and handling, the cylinder may consist of several segments that are assembled together during testing. However, in both design and assembly of the completed phantom one must insure a tight fit between adjacent segments, as even very small gaps will allow narrow axial regions of scatter-free radiation.

The test phantom line source insert is a clear polyethylene or polyethylene coated plastic tube at least 800 mm in length, with an inside diameter of 3.2 ± 0.2 mm (1/8") and an outside diameter of 4.8 ± 0.2 mm (3/16"). The central 700 ± 5 mm of this tube will be filled with a known quantity of activity and threaded through the 6.4 mm hole in the test phantom.

To begin the test, a source of relatively high activity is placed in the field of view of the positron emission tomograph. Regular measurements are then taken while the activity in the phantom decays over several half-lives. A decrease in the ratio of the random event rate to the true event rate accompanies the activity decay, eventually falling to less than 1%. In addition, the efficiency of the system in processing coincident events improves as the activity decays, until count losses may be effectively neglected. Thus by waiting long enough one obtains a measurement of the coincidence count rate that is effectively free from both randoms and processing losses. By
extrapolating this true rate back to higher activity levels and comparing it to the measured rate one may estimate count losses suffered by the system at higher activity levels. The accuracy of this technique depends critically on adequate statistics being gathered at sufficiently low levels of activity. This may require repeated measurements at the lower count rates.

![Figure 4-1: Positioning of Phantom](image)

**Figure 4-1**

**Positioning of Phantom**

### 4.3.1 Symbols

**Scatter fraction** (SF) — a dimensionless ratio of scattered coincidence events to the sum of scattered and true coincidence events in a defined ROI of the scanner field-of-view.

### 4.3.2 Radionuclide

The radionuclide used for this measurement shall be $^{18}$F. The amount of radioactivity shall be great enough to allow the following two rates to be measured:

a. $R_{\text{peak}}$ — peak true count rate

b. $R_{\text{NEC,peak}}$ — peak noise equivalent count rate

Recommendations for the initial activity required to meet these objectives will be supplied by the manufacturer.

The initial activity in the phantom shall be determined from the activity injected into the phantom as measured in a calibrated dose calibrator.

### 4.3.3 Source distribution

The central $700 \pm 5$ mm of the test phantom line source insert shall be filled with water well mixed with the measured amount of radioactivity and sealed at both ends. This line source shall be inserted into the hole of the test phantom such that the region of activity coincides with the $70$ cm length of the phantom. The test phantom with line source is mounted on the standard patient table supplied by the manufacturer and rotated such that the line source insert is positioned nearest to the patient bed (see Figure 4-1). The phantom is centered in the transverse and axial fields-of-view to within $5$ mm.

### 4.3.4 Data collection

Data shall be acquired at intervals more frequent than half the radionuclide half-life, $T_{1/2}$, until true events losses are less than 1.0%, and the random rates are less than 1.0% of true rates. The durations of the individual acquisitions, $T_{\text{acq}}$, shall be less than one-fourth of $T_{1/2}$. Acquisitions shall be fully tomographic; therefore, rotating scanners must rotate to provide complete and uniform angular sampling for each acquisition. In the case of rotating scanners, the acquisition time $T_{\text{acq}}$ shall include the time required to rotate the detectors.
It is essential to the accurate estimation of system dead-time losses that sufficient statistics be acquired with count loss rates and random rates both below 1.0% of true rates. Each acquisition should contain a minimum of 500,000 prompt counts. It is also important that the measurements around the peak count rate be done with sufficient frequency so that the peak rate can be accurately determined. Therefore, it is expected that manufacturers will recommend a protocol for their scanners that includes starting activity, acquisition times, and acquisition durations.

4.3.5 Data processing

For tomographs with an axial field of view of 65 cm or less, sinograms shall be generated for each acquisition $j$ of slice $i$. For tomographs with an axial field of view greater than 65 cm, sinograms shall be generated for each acquisition for slices within the central 65 cm. No corrections for variations in detector sensitivity or detector motions such as wobble, randoms, scatter, dead-time, or attenuation shall be applied to the measurements. Real-time subtraction of random events shall not be done.

The sinograms must contain the total acquired counts of the scanner without corrections; i.e., the sinograms must contain true, random, and scatter counts. Furthermore, scanners with the capacity for direct measurement of random rates may not use these measurements in the estimation of random count rates detailed below. Oblique sinograms are collapsed into a single sinogram for each respective slice (by single-slice rebinning) while conserving the number of counts in the sinogram.

4.4 ANALYSIS

All pixels in each sinogram $i$ of acquisition $j$ located farther than 12 cm from the center of the phantom shall be set to zero. For each projection angle $\phi$ within the sinogram, the location of the center of the line source response shall be determined by finding the pixel having the greatest value. Each projection shall be shifted so that the pixel containing the maximum value is aligned with the central pixel of the sinogram. After alignment, a sum projection shall be produced such that a pixel in the sum projection is the sum of the pixels in each angular projection having the same radial offset as the pixel in the sum projection:

$$C(r_{\lambda j}) = \sum_{a} C(r - r_{\max}(\phi), \phi)_{\lambda j}$$

Where:

a. $r$ is the pixel number in a projection,

b. $\phi$ is the projection number in the sinogram (i.e., the sinogram row), and

c. $r_{\max}(a)$ refers to the location of the maximum value in projection $\phi$.

The counts $C_{LU}$ and $C_{RU}$, the left and right pixel intensities at the edges of the 40 mm wide strip at the center of the sinogram, shall be obtained from the sum projection (see Figure 4-2). Linear interpolation shall be employed to find the pixel intensities at ± 20 mm from the central pixel of the projection. The average of the two pixel intensities $C_{LU}$ and $C_{RU}$ shall be multiplied by the number of pixels, including fractional values, between the edges of the 40 mm wide strip, and the product added to the counts in the pixels outside the strip, to yield the number of random plus scatter counts $C_{RSU}$ for the slice $i$ of acquisition $j$. 

The total event count $C_{TOT,i}$ is the sum of all pixels in the sum projection for slice $i$ of acquisition $j$. The average activity $A_{avg,i}$ for each acquisition $j$ shall be calculated (see Section 1.2).

4.4.1 Scatter fraction

The final acquisitions $j'$ of the sequence with count loss rates and random rates below 1.0% of the trues rate shall be used to determine the scatter fraction. For these acquisitions, it is assumed that $C_{r+i,b,i}$ has a negligible number of random counts and consists only of scatter counts, and likewise, $C_{TOT,i}$ consists only of true and scatter counts.

The scatter fraction $SF_i$ for each slice is calculated by summing over the low activity acquisitions as follows:

$$SF_i = \frac{\sum_{j} C_{r+s,i,j}^l}{\sum_{j} C_{TOT,i,j}^l}$$

The system scatter fraction $SF$ is computed as the weighted average of the $SF_i$ values as follows:

$$SF = \frac{\sum_{i} \sum_{j} C_{r+s,i,j}^l}{\sum_{i} \sum_{j} C_{TOT,i,j}^l}$$

4.4.2 Total event rate measurement

For each acquisition $j$, the total event rate $R_{TOT,i}$ for each slice $i$ is computed as:

$$R_{TOT,i} = \frac{C_{TOT,i}}{T_{acq,i}}$$

Where $T_{acq,i}$ is the acquisition time. The system total event rate $R_{TOT}$ is computed as the sum of $R_{TOT,i}$ over all slices $i$.

4.4.3 True event rate measurement

For each acquisition $j$, the true event rate $R_{i,j}$ for each slice $i$ is computed as:
\[ R_{Tli,j} = \frac{\left( C_{TOT,li,j} - C_{r+i,li,j} \right)}{T_{acq,j}} \]

Where \( T_{acq,j} \) is the acquisition time. The system true event rate \( R_{T,i} \) is computed as the sum of \( R_{Tli,j} \) over all slices \( i \).

### 4.4.4 Random event rate measurement

For each acquisition \( j \), the random event rate \( R_{r,i} \) for each slice \( i \) is computed as:

\[ R_{r,i,j} = R_{TOT,i,j} - \left( \frac{R_{Tli,j}}{1 - SF_i} \right) \]

The system random event rate \( R_{r,i} \) is computed as the sum of \( R_{r,i,j} \) over all slices \( i \).

### 4.4.5 Scatter event rate measurement

For each acquisition \( j \), the scatter event rate \( R_{s,i} \) for each slice \( i \) is computed as:

\[ R_{s,i,j} = \left( \frac{SF_i}{1 - SF_i} \right) R_{Tli,j} \]

The system scatter event rate \( R_{s,i} \) is computed as the sum of \( R_{s,i,j} \) over all slices \( i \).

### 4.4.6 Noise equivalent count rate measurement

For each acquisition \( j \), on all systems except those which perform direct randoms subtraction, the noise equivalent count rate \( R_{NEC,i} \) for each slice \( i \) is computed as:

\[ R_{NEC,i,j} = \frac{R_{Tli,j}^2}{R_{TOT,i,j}} \]

Systems that use direct randoms subtraction should instead compute \( R_{NEC,i,j} \) for each slice \( i \) as:

\[ R_{NEC,i,j} = \frac{R_{Tli,j}^2}{R_{TOT,i,j} + R_{r,i,j}} \]

The system noise equivalent count rate \( R_{NEC,i} \) is computed as the sum of \( R_{NEC,i,j} \) over all slices \( i \).

### 4.5 REPORT

#### 4.5.1 Count rate plot

For the system, plot the following five quantities as a function of the average effective radioactivity concentration \( a_{ave,j} \) as defined in Section 1.2, where the volume \( V \) is the total volume of the cylindrical phantom (22,000 cm\(^3\)):

a. \( R_{T,i} \) — system true event rate  
b. \( R_{r,i} \) — system random event rate  
c. \( R_{s,i} \) — system scatter event rate 
d. \( R_{NEC,i} \) — system noise equivalent count rate 
e. \( R_{TOT,i} \) — system total event rate
4.5.2 Peak count values

Report the following values, derived from the above plot:

a. $R_{t,\text{peak}}$  – peak true count rate
b. $R_{\text{NEC,peak}}$ – peak noise equivalent count rate
c. $a_{t,\text{peak}}$  – the activity concentration at which $R_{t,\text{peak}}$ is reached
d. $a_{\text{NEC,peak}}$ – the activity concentration at which $R_{\text{NEC,peak}}$ is reached

4.5.3 System scatter fraction

Report the value of $SF$. 
APPENDIX B
HOW TO COPY SPECIFIC FILES TO A CD

- Go to **File Management**
- **Double click** a patient’s name
- Press and hold **control** key, then click with the mouse on the desired files to highlight the files you want to copy (usually ----.scn). Add up the sizes of the files picked to make sure they add up to less than 700 MB (CD capacity)
- Go to **File**, then **Copy File (s)**. A window will pop up
- Type in the directory to which the files will be copied: /sun0/patient/dicom. This will take a minute
- Open an x-term window
- Type: **cd** /sun0/patient/dicom
- Then type: **ls**. This will list the files you copied to the dicom directory.
- Right click with the mouse on the background, just as you would do to open an x-term window.
- Go to **Utilities**, **Makeimage**, then **Burn CD**. You’re done until this point
- However, you will need to empty the dicom directory if you want to burn more files again. So, go back to **Utilities** and click **Clean DICOM Directory**.
APPENDIX C
A SHELL SCRIPT FOR SCALING FACTORS

This program extracts the scaling factors information found in the subheaders of each sinogram. To run the program, the following commands should be used. Notice that the name of this file is scnscl.sch and therefore the commands used refer to this file. The first command to be used is `chmod 555 scnscl.sch`, followed by `./scnscl.sch filename.scn 2`. The filename here refers to the Fourier rebinned file. The output from this program will consist of three columns indicating frame number, slice index, and scaling factor, respectively.

```csh
#!/bin/csh -f
# First parameter is name of .scn file.
#
# The second parameter is number of first slice in each frame.
#
set scnfilename = $1
set firstslice  = $2

set nslices = 45
set nframes = 21
set slicethickness = 4

set i = 1
set f = 1

while ( $f <= $nframes )
  while ( $i <= $nslices )
    @ s = ( $i - 1 ) * $slicethickness + $firstslice
    echo -n $f $s " "
    sbhdrdmp $1 $s 0 $f | fgrep scnscl | awk '{ print $4 }'
    @ i = $i + 1
  end
  @ f = $f + 1
  set i = 1
end
```

50
%***************************************************************************************************
% AQUISITIONS CODE
%***************************************************************************************************

% The acquisitions' raw data acquired on the PET scanner are all saved in one huge
% scan file (xxx.scn). The interpolated scan file (xxx_int.scn) is generated by the
% reconstruction program on the scanner. This interpolated file has 45 slices per tilt
% angle. There are 7 tilt angles per acquisition. Moreover, 192 projection angles and
% 256 linear samples are present for each acquisition. Notice that the Fourier Rebinned
% file doesn't have any tilt angles involved.

% This program extracts the sinograms from the scan file. The extracted sinograms of
% each acquisition are output to different acquisition files numbered 1 through 21. The
% main header in the file, the subheader of each sinogram, and the directory record
% header are all extracted and are not included in the output acquisition files.

fid=fopen('p311s0_cou_FORE.scn','rb','ieee-be')

% Input parameters
nray=256; nang=192; nslc=45; nacq=21;
mainH=512; %main header
subH=512; %sub header
DR=1:1:31; %directory records
Tnslc=945; %Total number of slices

% Reading sinograms from file into scn matrices
j=1; k=1;
for i=1:1:Tnslc
    nbytes_skip=mainH+DR(j)*512+subH+(nbytes+subH)*sinog_num(i); % # of bytes to skip
    status=fseek(fid,nbytes_skip,-1); % returns 0 if success
    position=ftell(fid); % tells indicator position in the file
scn(:,:,i)=fread(fid,[nray nang],'int16');
if (i == j*31) %to add bytes of new directory record
    j=j+1;
end
end

%-------------------------------------------------------------------------------------------------------------------
% Output acquisitions only. These acquisitions are free of any header information
%-------------------------------------------------------------------------------------------------------------------
acquisition=zeros(nray,nang);
START=1; END=45;
for acq=1:1:nacq
    fname=sprintf('acquisitions\acq%d',acq);
    fid=fopen(fname,'wb','ieee-be');
    for i=START:1:END
        acquisition=scn(:,:,i);
        fwrite(fid,acquisition(:),'int16');
        acquisition=zeros(nray,nang);
    end
    fclose(fid)
    START=START+45;
    END=END+45;
end

%**************************************************************************************************
%                                                          END OF CODE                                                    *
%**************************************************************************************************
APPENDIX E
MATLAB CODE FOR SCALING FACTORS

%***************************************************************************************************
%*                                             SCALING FACTORS CODE                                             *
%***************************************************************************************************
%* This program organizes the scaling factors in a format suitable for
%* processing in the Data Processing Code.
%***************************************************************************************************
load ('sclfact.txt')

global ScalingFactors

k=1;
for j=1:1:26
    for i=1:1:45
        ScalingFactors(i,j)=sclfact(k,3);
        k=k+1;
    end
end

clear

%***************************************************************************************************
%*                                            END OF CODE                                            *
%***************************************************************************************************
APPENDIX F
MATLAB CODE FOR DATA PROCESSING AND ANALYSIS

%**************************************************************************************************
%* The source files used in this program contain data acquired from the dedicated
%* 3-D PET scanner in the Radiology Department at Shands Medical Plaza at the
%* University of Florida. This program processes the acquired data according to
%* section 4 of NEMA 2001 Standards.
%**************************************************************************************************

%**************************************************************************************************
%*                           NEMA NU 2-2001 DATA PROCESSING CODE
%**************************************************************************************************

global ScalingFactors
global FinalSumProj
global nray nang nslcs nacq
global mm_pxl cm_pxl CenterPixel pxl_L pxl_R
global sino ssino
global xx_R xx_L npxl_fourcm x_cm
global ScatF TRandScat TCounts

%-------------------------------------------------------------------------------------------------------------------
% Input Parameters describing the sinogram size and the number of acquisitions involved
%-------------------------------------------------------------------------------------------------------------------
nray=256; % number of linear samples or rays
nang=192; % number of projection angles
nslcs=45; % number of slices in an acquisition
nacq=21; % number of acquisitions
mm_pxl=576/256; % sampling distance (i.e., # of mm per pixel)
%-------------------------------------------------------------------------------------------------------------------
CenterPixel=nray/2;
cm_pxl=mm_pxl/10; % cm per pixel
pxl_twelvecm=12/cm_pxl; % # of pixels in 12 cm
pxl_twelvecm=ceil(pxl_twelvecm); % rounds the number to the next integer
pxl_L=CenterPixel+pxl_twelvecm-1; % pixel # located 12 cm to the left side of the center
% of the phantom
pxl_R=CenterPixel+pxl_twelvecm+1; % pixel # located 12 cm to the right side of the
% center of the phantom
%-------------------------------------------------------------------------------------------------------------------
xx_R=round(CenterPixel + 20/mm_pxl); % pixel # located 2 cm to the right from the Central pixel
xx_L=round(CenterPixel - 20/mm_pxl); % pixel # located 2 cm to the left from the central pixel
npxl_fourcm=ceil(40/mm_pxl); % # of pixels in 4 cm (40 mm) strip
% Converting x-axis to cm scale
ncm=CenterPixel*cm_pxl;  % number of cm from pixel 1 to central pixel (128)
x_cm=-(ncm-cm_pxl):cm_pxl:ncm;  % new scale in cm

sino=zeros(256,192,45);  %initializing zero matrices for the sinograms
ssino=zeros(256,192,45);  %initializing zero matrices for the shifted sinograms
FinalSumProj=zeros([1 nray nacq]);  %initializing a zero matrix for the Final Sum Projection

for acq=1:1:nacq
  global ScalingFactors
  global FinalSumProj
  global nray nang nslcs
  global mm_pxl cm_pxl CenterPixel pxl_L pxl_R
  global sino ssino
  global xx_R xx_L npxl_fourcm x_cm
  global ScatF TRandScat TCounts

  % open and read from each acquisition
  disp('opening acquisition'),disp(acq)  % to keep track of the code execution process
  fname=sprintf('acquisitions\acq%d',acq);
  fid=fopen(fname,'rb','ieee-be');

  % Notice: the 'ieee-be' defines the machine format used to read the acquisitions data.
  % Here the data was acquired at a Unix Sun Computer, so the data is saved as big
  % endian which is the format associated with Unix Operating Systems. PC or Windows
  % Operating Systems, on the other hand, are of little endian type. Therefore, when
  % reading Unix data on a PC it is essential to define the machine format before reading
  % the data. This was found to be better than swapping the data before using it in PC
  % environment, because the swapping process might introduce some unwanted
  % data embedded within the original data.
  disp('reading sinograms')  % to keep track of the code execution process
  slc_num=0:1:(nslcs-1);  % # slices to skip
  I=zeros(nray,nang,nslcs);  % Initializing zero matrices
  for i=1:1:nslcs
    NumBytesSkip=(nray*nang*2)*slc_num(i);  %number of bytes to skip
    status=fseek(fid,NumBytesSkip,-1);  %returns 0 if success
    position=ftell(fid);  %tells indicator position in the file
    I(:,:,i)=fread(fid,[nray,nang],'int16');  %sinogram images
  end
  fclose(fid);
% Multiplying by the scaling factors to correct for the number of counts in each pixel of each sinogram
%-------------------------------------------------------------------------------------------------------------------

```
disp('multiplying by the scaling factors') % to keep track of the code execution process
for i=1:1:nslcs
    I(:,:,i)=ScalingFactors(i,acq).*I(:,:,i);
end

disp('open and read done') % to keep track of the code execution process
```

%==================================================================
% 4.4 ANALYSIS (NEMA 2001)
%==================================================================
% Calculating the sum projection of each slice i in each acquisition j and the final sum projection of each acquisition.
%-------------------------------------------------------------------------------------------------------------------

```
disp('Analysis begins')
```

% Initialization
```
u=1; % index for ray
sinogram=zeros([nray nang]);
shiftedsinogram=zeros([nray nang]);
shiftedsinogramtranspose=zeros([nang nray]);
MaxCount=zeros([1 nray]);
ray=zeros;
angle=zeros;
SumProj=zeros([1 nray nslcs]);
finalsumprojection=zeros([1 nray]);
```

```
for i=1:1:nslcs     % Notice that there are 2 mid-points, 128 & 129 pxls
    sino(:,:,i)=I(:,:,i); % saves all sinograms of one acquisition only
    sinogram=I(:,:,i);
    sinogram(1:pxl_L,:)=0; % pxls 12cm to the left from center of phantom are zeroed.
    sinogram(pxl_R:nray,:)=0; % pxls 12cm to the right from center of phantom are zeroed.

    %---------------- ------------------------------------------------------------------------------------------
    % Finding the pixel of maximum count and the location of that pixel within the sinogram slice
    %---------------- ------------------------------------------------------------------------------------------
    MaxCount=max(sinogram); % returns max number of counts in each proj angle
    for phi=1:1:nang
        if (MaxCount(phi) == 0)
            continue;
        end
    end
```
elseif (MaxCount(phi) > 0)
    r=find (sinogram(:,phi)==MaxCount(phi)); % Tells location of PixelMaxCount.
    % Returns the row pixel location.
    if(length(r)==1) % if only one value returned (ideal case)
        ray(u)=r;
    elseif(length(r)>1) % if more than one value returned
        ray(u)=min(r);
        % Notice that in the same projection angle there
        % may be more than one pixel that has max counts.
        % Only the nearest pixel to the center of the line
        % source is taken as the pixel of maximum count.
    end
    angle(u)=phi; % tells location of PixelMaxCount
    % returns the column pixel location=projection angle
    u=u+1; % increment index u
end
% check if sizes of row and clmn arrays are the same
if(size(ray) ~= size(angle))
    'sizes not equal'
end
%---------------------------------------------------------------------------------------------------
% Shifting the pixels around until the pixel of maximum count aligns with the
%acentral pixel (i.e., pixel # 128)
%---------------------------------------------------------------------------------------------------
for j=1:1:length(angle)
    ProjAngle=angle(j);
    PixelMaxCount=ray(j);
    offset=CenterPixel-PixelMaxCount;
    for r=(pxl_L+1):1:(pxl_R-1)
        r_new=r+offset;
        if(r_new < 1)
            r_new=1;
        elseif(r_new > nray)
            r_new=nray;
        end
        shiftedsinogram(r_new,ProjAngle)=sinogram(r,ProjAngle);
    end
end
%This is based on zeroing of the array before and assumes that
%the difference is less than 79 pixels or 17.78 cm
shiftedsinogram(1:48,:)=0;
shiftedsinogram(208:nray,:)=0;
ssino(:,:,i)=shiftedsinogram;
%---------------------------------------------------------------------------------------------------
% Saving all important data each in the appropriate matrix
% ShiftedSinogram(:,:,i)=shiftedsinogram; % shifted proj, result is straight line
shiftedsinogramtranspose=shiftedsinogram';
SumProj(:,:,i)=sum(shiftedsinogramtranspose); % 1-D profile 'sum in every column'
finalsumprojection=finalsumprojection+SumProj(:,:,i);

% Calculating the scatter + Random counts and the total counts per slice

y=zeros([1 nray]);
y=SumProj(:,:,i);
C_R=y(xx_R);
C_L=y(xx_L);
avg=(C_R+C_L)/2;

sum_R=sum(y((xx_R+1):nray)); % counts in pixels outside the strip
sum_L=sum(y(1:(xx_L-1))); % counts in pixels outside the strip
sum_RL=sum_R+sum_L; % total counts in pixels outside the strip

C_rand_scat=(avg*npxl_fourcm)+sum_RL;
C_r_s_j(i)=C_rand_scat; % random plus scatter counts for acquisition j
C_TOT_j(i)=sum(y); % total counts per acquisition j

% re-zeroing the matrices to avoid data overlap
sinogram=zeros([nray nang]); % zero out the sinogram matrix
shiftedsinogram=zeros([nray nang]); % zero out the shifted sinogram matrix
shiftedsinogramtranspose=zeros([nang nray]);
MaxCount=zeros([1 nang]);
ray=zeros;
angle=zeros;
r=zeros;
u=1; % index for row
end % closes the nslices loop
disp('Calculating Random and Scatter, and Total counts')
TRandScat(acq)=sum(C_r_s_j); % Total Randoms and scatters in an acq
TCounts(acq)=sum(C_TOT_j); % Total counts in an acq
ScatF(acq)=TRandScat(acq)/TCounts(acq);

% output counts per acquisition to a text file
SliceNum=1:1:45;
Data1=[SliceNum;C_r_s_J;C_TOT_J];
Data2=[TRandScat(acq);TCounts(acq)];
fname=sprintf('Output\DataCounts%d.mw',acq);
fid=fopen(fname,'w');
fprintf(fid,'acq ');
fprintf(fid,'%2.0f
',acq);
fprintf(fid,'Slice	 RandScat	 Total
');
fprintf(fid,'%2.0f	 %10.2f	 %10.2f
',Data1);

fprintf(fid,'

');
fprintf(fid,'Total 	 %10.2f %10.2f',Data2);
close(fid);

%--------------------------------------------------------------------------------------------------------------
% In slight deviation from NEMA 2001, the sum projections of all sinograms in a
% particular acquisition are summed to create a final sum projection that describes
% the behavior of the PET scanner at that particular activity. It is this final sum
% projection that is used to calculate the different count rates following NEMA 2001
% standards.
%--------------------------------------------------------------------------------------------------------------
FinalSumProj(:,:,acq)=finalsumprojection; % final sum projection of each acquisition

%---------------------------------------------------------------------------------------------------------------
% output Final Sum Projections data to a text file
%---------------------------------------------------------------------------------------------------------------
T_acq_j=20*60;
Data3=[x_cm;FinalSumProj(:,:,acq)/T_acq_j];

fname=sprintf('Output\FinalSumProjection%d.mw',acq);
fid=fopen(fname,'w');
fprintf(fid,'acq ');
fprintf(fid,'%2.0f
',acq);
fprintf(fid,'x (cm)	 Count Rate [cps] 
');
fprintf(fid,'%2.3f	 %10.2f 
',Data3);
close(fid);

clear %clears all but the global variables
end %ends the nacq loop

global cm_pxl x_cm CenterPixel
global nacq FinalSumProj

%---------------------------------------------------------------------------------------------------------------
% Plot Final Sum Projections (cps) vs. radial distance (cm)
%---------------------------------------------------------------------------------------------------------------
T_acq_j=20*60;
% N=1; fig=1;
% for j=1:1:nacq
%   fname=sprintf('Final Sum Projection: acq%d.j');
%   figure(fig),subplot(2,1,N),plot(x_cm,FinalSumProj(:,:,j)/T_acq_j),title(fname)
%   xlabel('Distance along transverse axis [cm]'),ylabel('Count Rate [cps]')
%   grid
%   N=N+1;
%   if(j==fig*2)
%       fig=fig+1;
%   N=1;
%   end
% end
%-----------------------------------------------------------------------------------------------------------------
% To place each profile in a separate figure
%-----------------------------------------------------------------------------------------------------------------
fig=1;
for j=1:1:nacq
    figure(fig),plot(x_cm,FinalSumProj(:,:,j)/T_acq_j)
    xlabel('Distance along transverse axis [cm]'),ylabel('Count Rate [cps]')
    grid
    fig=fig+1;
end
%**************************************************************************************************
%*                                                         END OF CODE                                                    *
%**************************************************************************************************
APPENDIX G
MATLAB CODE FOR CONTINUING PROCESSING AND ANALYSIS

%**************************************************************************************************
%*                   NEMA NU 2-2001 DATA PROCESSING CODE (Continue ...)                 *
%**************************************************************************************************
%global FinalSumProj mm_pxl
%global ScatF TRandScat TCounts

T_acq_j=20*60; %acquisition time in seconds
C_r_s_j=TRandScat;
C_TOT_j=TCounts;

%------------------------------------------------------------------------------------------------------------------
% 4.4.1 SCATTER FRACTION (NEMA 2001)
%------------------------------------------------------------------------------------------------------------------
% The final acquisition j' of the sequence with count loss rates and random rates below
% 1% of the trues rate is used to determine the scatter fraction. For this acquisition, it is
% assumed that C_r+s,j' has a negligible number of random counts and consists only of
% scatter counts, and likewise, C_TOT,j' consists only of true and scatter counts. The
% scatter fraction of the final acquisition is used as the system's SF for it is assumed
% that the randoms rate is again negligible in that acquisition and therefore the C_r+s,j'
% consists only of scatter counts.
%------------------------------------------------------------------------------------------------------------------

%------------------------------------------------------------------------------------------------------------------
% 4.5.3 SYSTEM SCATTER FRACTION (NEMA 2001)
%------------------------------------------------------------------------------------------------------------------
SF =0.403;

%------------------------------------------------------------------------------------------------------------------
% 4.4.2 TOTAL EVENT RATE MEASUREMENT (NEMA 2001)
%------------------------------------------------------------------------------------------------------------------
R_TOT_j = C_TOT_j ./ T_acq_j;

%------------------------------------------------------------------------------------------------------------------
% 4.4.3 TRUE EVENT RATE MEASUREMENT (NEMA 2001)
%------------------------------------------------------------------------------------------------------------------
R_t_j = (C_TOT_j - C_r_s_j) ./ T_acq_j;

%------------------------------------------------------------------------------------------------------------------
% 4.4.4 RANDOM EVENT RATE MEASUREMENT (NEMA 2001)
%------------------------------------------------------------------------------------------------------------------
R_r_j = R_TOT_j - (R_t_j ./ (1-SF));
% 4.4.5 SCATTER EVENT RATE MEASUREMENT (NEMA 2001)
R_s_j = R_t_j .*(SF/(1-SF));

% 4.4.6 NOISE EQUIVALENT COUNT RATE MEASUREMENT (NEMA 2001)
R_NEC_j = (R_t_j.^2) ./ R_TOT_j;

% 4.5 REPORT
% 4.5.1 COUNT RATE PLOT (NEMA 2001)
% calculated initial activity (mCi) at the beginning of each acquisition is
Act=[12.536 9.738 7.564 5.876 4.564 3.545 2.754 2.139 1.662 1.291 1.003 0.779 0.605 …
     0.470 0.365 0.284 0.220 0.171 0.133 0.103 0.080]; % mCi;

% time into experiment (minutes):
time=[0 40 80 120 160 200 240 280 320 360 400 440 480 520 560 600 640 680 720 760 800];

figure(1),
plot(time,R_TOT_j/1000,'m*-',time,R_t_j/1000,'rs-',time,R_r_j/1000,'bd-',time,R_s_j/1000,'gx-')
xlabel('Time [min]'),ylabel('Count Rate [kcps]')
legend('Total','True','Random','Scatter')

figure(2)
plot(Act,R_TOT_j/1000,'m*-',Act,R_t_j/1000,'rs-',Act,R_r_j/1000,'bd-',Act,R_s_j/1000,'gx-')
xlabel('Activity [mCi]'),ylabel('Count Rate [kcps]')
legend('Total','True','Random','Scatter')

figure(3)
plot(time,R_NEC_j/1000,'k.-')
xlabel('Time [min]'),ylabel('NEC Count Rate [kcps]')

figure(4)
plot(Act,R_NEC_j/1000,'k.-')
xlabel('Activity [mCi]'),ylabel('NEC Count Rate [kcps]')

% output count rates to a text file
Data=1:1:nacq;
countdata = [Data; C_TOT_j; R_TOT_j; R_t_j; R_r_j; R_s_j; R_NEC_j; Act];

fid=fopen('Output\CoincidenceCounts.txt','w')
fprintf(fid,'Acquisition  Total Paired  Count Rate  Count Rate  Count Rate  Count Rate  Count Rate  Count Rate  Average\n')
fprintf(fid,'                Events          Total          True         Random       Scatter       NEC      Activity\n')
fprintf(fid,'             (counts/20min)     (cps)          (cps)         (cps)        (cps)       (cps)       (mCi) \n')
fprintf(fid,'%2.0f          %12.0f   %12.2f   %12.2f  %12.2f  %12.2f  %10.2f    %1.4f
',countdata)
close(fid);

%**************************************************************************************************
%*                                                          END OF CODE                                                   *
%**************************************************************************************************

%**************************************************************************

%*                      END OF CODE                                        *

%**************************************************************************
%***************************************************************************************************
%*                                                      CURVE FIT CODE                                                   *
%***************************************************************************************************
%-------------------------------------------------------------------------------------------------------------------
% Fitting the Final Sum Projections to Gaussian Functions using the Non-Linear Least Squares Method
%-------------------------------------------------------------------------------------------------------------------

% Fitting the Final Sum Projections to Gaussian Functions using the Non-Linear Least Squares Method

% global FinalSumProj
% global nacq cm_pxl CenterPixel
% T_acq_j=20*60;

ncm=CenterPixel*cm_pxl; % number of cm from pixel 1 to central pixel (128)
x_cm=-(ncm-cm_pxl):cm_pxl:(ncm); % new scale in cm

for j=1:1:nacq
    Total(:,:,j)=FinalSumProj(:,:,j)/T_acq_j;
end

model = fittype('a*exp(-0.5*(x-b)^2/c^2)+d'); % the gaussian equation used to fit the data
opts  = fitoptions('Method','NonlinearLeastSquares');
opts.Lower = [-Inf -Inf 0 -Inf];
for j=1:1:nacq
    a_strt=max(Total(:,:,j)); % Normalizes the gaussian function
    b_strt=1; % the mean
    c_strt=1; % the standard deviation
    d_strt=1; % the offset
    opts.StartPoint=[a_strt b_strt c_strt d_strt]; % starting points of statistical constants
    [f,gof] = fit(x_cm',Total(:,:,j)',model,opts);
    a(j)=f.a;
b(j)=f.b;c(j)=f.c;d(j)=f.d;
R_squared(j)=gof.rsquare;
G{j}=f; % Gaussian fitted equation
j
end
%
% Calculating the FWHM of the fitted curves. FWHM = 2.35 * sigma
%
for j=1:1:nacq
    FWHM(j)=c(j)*2.35*10; % FWHM (mm)
    FWHMa(j)=sqrt(FWHM(j)^2-3.2^2); %adjusted FWHM (mm)
end
%
% Output the FWHM values to a text file
%
acqs=1:1:nacq;
values=[acqs; FWHM; FWHMa];
fid=fopen('Output\FWHM.txt','w');
fprintf(fid,'acq	 FWHM(mm)	 FWHMa(mm)	 
');
fprintf(fid,'%2.0f	 %3.2f	 %3.2f
',values);
fclose(fid);
%
% Output the coefficients values, for each fitted gaussian function, to a text file
%
coeff=[a; b; c; d; R_squared];
fid=fopen('Output\FittedFunctionsCoeff.txt','w');
fprintf(fid,'	 a		 b		 c		 d	 R_squared 
');
fprintf(fid,'%7.2f  %1.4f  %1.5f  %7.2f  %1.3f 
',coeff);
fclose(fid);
%
% Output fitted curves to a text file
%
for j=1:1:nacq
    f=G{j}; %fitted gaussian equation
    for i=1:1:length(x_cm)
        fg(i)=f(x_cm(i));
    end
    g=[x_cm;fg];
    fname=sprintf('Output\FittedGaussian%d.mw',j);
    fid=fopen(fname,'w');
    fprintf(fid,'acq 
');
    fprintf(fid,'%2.0f
',j);
    fprintf(fid,'x (cm)	 Count Rate [cps] 
');
    fprintf(fid,'%2.3f	 %10.2f
',g);
    fclose(fid);
end
%
% Plot the FWHM as a function of activity and time into experiment
%
figure(30),plot(Act,FWHMa)
xlabel('Activity [mCi]'), ylabel('FWHM [mm]')
figure(31), plot(time, FWHMa)
xlabel('Time [min]'), ylabel('FWHM [mm]')

%-------------------------------------------------------------------------------------------------------------------
% Plotting the fitted functions along with the Final Sum Projections
%-------------------------------------------------------------------------------------------------------------------
% N=1;
% fig=1;
% for j=1:1:nacq
%     fname=sprintf('Final Sum Projection: Acq%d',j);
%     figure(fig), subplot(2,1,N), plot(G{j}, x_cm, Total(:,:,j)), title(fname)
%     xlabel('Distance along transverse axis [cm]'), ylabel('Counts')
%     legend('Final Sum Projection','Gaussian'), grid
%     N=N+1;
%     if(j==fig*2)
%         fig=fig+1;
%         N=1;
%     end
% end

fig=1;
for j=1:1:nacq
    figure(fig), plot(G{j}, x_cm, Total(:,:,j))
    xlabel('Distance along transverse axis [cm]'), ylabel('Count Rate [cps]')
    legend('Final Sum Projection','Fitted Gaussian'), grid
    fig=fig+1;
end

%-------------------------------------------------------------------------------------------------------------------
% Fitting the Randoms rate curve to power equation using the Non-Linear Least Squares Method
%-------------------------------------------------------------------------------------------------------------------
model = fittype('power1');
opts  = fitoptions('Method','NonlinearLeastSquares');
opts.Lower = [-Inf -Inf];
opts.Upper = [ Inf 2];
opts.StartPoint=[1 2];
% R_r_j=R_r_j/1000; % kcps
[Rand, gof, out] = fit(Act', R_r_j', model, opts);
%-------------------------------------------------------------------------------------------------------------------
% Output fitted randoms curve to a text file
%-------------------------------------------------------------------------------------------------------------------
for i=1:1:length(Act)
    R(i)=Rand(Act(i));
end
r=[Act;R];
fid=fopen('Output\FittedRandoms.mw','w');
fprintf(fid,'Act (mCi)\t Count Rate [kcps] \n');
fprintf(fid,'%2.3f\t %10.2f\n',r);
fclose(fid);

%-------------------------------------------------------------------------------------------------------------------
% Plotting the fitted function along with the Randoms count rate
%-------------------------------------------------------------------------------------------------------------------
figure(fig),plot(Rand,Act,R_r_j)
%title('Random count rate and fitted equation vs. activity')
xlabel('Activity [mCi]'), ylabel('Count Rate [cps]')
legend('Randoms rate','Fitted equation'),grid
%***************************************************************************************************
%*                                                           END OF CODE                                                   *
%***************************************************************************************************
APPENDIX I
MATLAB CODE FOR PRODUCING MTF CURVES

%*************************************************************************
%* Modulation Transfer Function (MTF) CODE                              *
%*************************************************************************
%* The Fourier Transform of the Final Sum Projection is computed to produce the
%* Modulation Transfer Function (MTF) for each acquisition.              *
%*************************************************************************

% Fourier Transform of Final Sum Projections
%-------------------------------------------------------------------------------------------------------------------

global nray

MTF = zeros([1 nray nacq]); % Modulation Transfer Function
MTFn = zeros([1 nray nacq]); % Normalized Modulation Transfer Function
MTFa = zeros([1 nray/2 nacq]); % Adjusted Modulation Transfer Function
for j = 1:nacq
    fsp = FinalSumProj(:,:,j); % fsp = final sum projection
    FSP = fftshift(fft(fsp));
    MTF(:,:,j) = abs(FSP); % magnitude of FT{fsp}
end
%-------------------------------------------------------------------------------------------------------------------

% Normalized Modulation Transfer Functions (MTFn)
%-------------------------------------------------------------------------------------------------------------------

for j = 1:nacq
    MAX = max(MTF(:,:,j));
    MTFn(:,:,j) = MTF(:,:,j)/MAX;
end
%-------------------------------------------------------------------------------------------------------------------

% Adjusted Modulation Transfer Functions (MTFa)
%-------------------------------------------------------------------------------------------------------------------

for j = 1:nacq
    MTFa(:,:,j) = MTFn(:,:,nray/2 + 1:nray,j);
end
%-------------------------------------------------------------------------------------------------------------------

% Plotting the adjusted Modulation Transfer Functions (MTFa)
%-------------------------------------------------------------------------------------------------------------------

N = nray % # of pixels, so 1mm/pixel
dx = cm_pxl % sampling distance: i.e., # of cm per pixel
dnu = 1/(N*dx) % sampling distance in frequency domain
nu_min=-(N*dnu/2)
nu_max=(N*dnu/2)-dnu

nu=nu_min:dnu:nu_max; % frequency domain (1/cm)
nu1=nu((nray/2 + 1):nray); % frequency domain used for the plots

% fig=1;
% subfig=1;
% for j=1:1:nacq
%   figure(fig),subplot(2,1,subfig),plot(nu1,MTFa(:,:,j))
%   title([‘Modulation Transfer Function: Acq ’ int2str(j)])
%   xlabel(‘nu [cm^-1]’),ylabel(‘Gain’),grid
%   subfig=subfig+1;
%   if(j==fig*2)
%     fig=fig+1;
%     subfig=1;
%   end
% end
% fig=1;
for j=1:1:nacq
    figure(fig),plot(nu1,MTFa(:,:,j))
    xlabel(‘nu [cm^-1]’),ylabel(‘Gain’)
    h = gca;
    set(h,’YGrid’,’on’);
    fig=fig+1;
end

%-------------------------------------------------------------------------------------------------------------------
% Plotting the adjusted Modulation Transfer Functions (MTFa) of selected acquisitions
% on ONE plot
%-------------------------------------------------------------------------------------------------------------------
figure(20)
plot(nu1,MTFa(:,:,1),nu1,MTFa(:,:,5),nu1,MTFa(:,:,10),nu1,MTFa(:,:,21))
xlabel(‘nu [cm^-1]’),ylabel(‘Gain’)
h = gca;
set(h,’YGrid’,’on’);
%title(’Modulation Transfer Functions of selected acquisitions’)
legend(’acquisition 1’,’acquisition 5’,’acquisition 10’,’acquisition 21’)

%-------------------------------------------------------------------------------------------------------------------
% Output MTF curves to text files
%-------------------------------------------------------------------------------------------------------------------
for acq=1:1:nacq
    mtf=[nu1;MTFa(:,:,acq)];
    fname=sprintf(‘Output\MTF%d.mw’,acq);
    fid=fopen(fname,’w’);
    fprintf(fid,’acq ‘);
    fprintf(fid,’%2.0f
’,acq);
    fprintf(fid,’frequency (1/cm)\t Gain \n’);
fprintf(fid,'%2.3f	%10.5f
',mtf);
fclose(fid);
end

%-------------------------------------------------------------------------------------------------------------------
% Plotting ALL of the adjusted Modulation Transfer Functions (MTFa) on ONE plot
%-------------------------------------------------------------------------------------------------------------------
% for j=1:1:nacq
%     ModTranFuna(j,:)=MTFa(:,:,j);
%     fname=sprintf('s%d',j);
%     ACQMTFa{j}=fname;
% end
% figure(21)
% plot(nu1,ModTranFuna),legend(ACQMTFa)
%-------------------------------------------------------------------------------------------------------------------
% Plotting ALL of the Modulation Transfer Functions (MTF) on ONE plot
%-------------------------------------------------------------------------------------------------------------------
% for j=1:1:nacq
%     ModTranFunNorm(j,:)=MTFn(:,:,j);
%     flname=sprintf('s%d',j);
%     ACQMTFn{j}=flname;
% end
% figure(22)
% plot(nu,ModTranFunNorm),legend(ACQMTFn)
% xlabel('nu [cm^-^1]'),ylabel('Relative intensity')
% title('Norm MTFs')
%******************************************************************************
%*                                                                                       END OF CODE
%******************************************************************************
APPENDIX J
FINAL SUM PROJECTIONS

FIG. 8. Final sum projection of acquisition 1
FIG. 9. Final sum projection of acquisition 2

FIG. 10. Final sum projection of acquisition 3
FIG. 11. Final sum projection of acquisition 4

FIG. 12. Final sum projection of acquisition 5
FIG. 13. Final sum projection of acquisition 6

FIG. 14. Final sum projection of acquisition 7
FIG. 15. Final sum projection of acquisition 8

FIG. 16. Final sum projection of acquisition 9
FIG. 17. Final sum projection of acquisition 10

FIG. 18. Final sum projection of acquisition 11
FIG. 19. Final sum projection of acquisition 12

FIG. 20. Final sum projection of acquisition 13
FIG. 21. Final sum projection of acquisition 14

FIG. 22. Final sum projection of acquisition 15
FIG. 23. Final sum projection of acquisition 16

FIG. 24. Final sum projection of acquisition 17
FIG. 25. Final sum projection of acquisition 18

FIG. 26. Final sum projection of acquisition 19
FIG. 27. Final sum projection of acquisition 20

FIG. 28. Final sum projection of acquisition 21
APPENDIX K
FITTED FINAL SUM PROJECTIONS

FIG. 36. Final sum projection of acquisition 1 and its fitted gaussian curve

FIG. 37. Final sum projection of acquisition 2 and its fitted gaussian curve
FIG. 38. Final sum projection of acquisition 3 and its fitted gaussian curve

FIG. 39. Final sum projection of acquisition 4 and its fitted gaussian curve
FIG. 40. Final sum projection of acquisition 5 and its fitted gaussian curve

FIG. 41. Final sum projection of acquisition 6 and its fitted gaussian curve
FIG. 42. Final sum projection of acquisition 7 and its fitted gaussian curve

FIG. 43. Final sum projection of acquisition 8 and its fitted gaussian curve
FIG. 44. Final sum projection of acquisition 9 and its fitted gaussian curve

FIG. 45. Final sum projection of acquisition 10 and its fitted gaussian curve
FIG. 46. Final sum projection of acquisition 11 and its fitted gaussian curve

FIG. 47. Final sum projection of acquisition 12 and its fitted gaussian curve
FIG. 48. Final sum projection of acquisition 13 and its fitted gaussian curve

FIG. 49. Final sum projection of acquisition 14 and its fitted gaussian curve
FIG. 50. Final sum projection of acquisition 15 and its fitted gaussian curve

FIG. 51. Final sum projection of acquisition 16 and its fitted gaussian curve
FIG. 52. Final sum projection of acquisition 17 and its fitted gaussian curve

FIG. 53. Final sum projection of acquisition 18 and its fitted gaussian curve
FIG. 54. Final sum projection of acquisition 19 and its fitted gaussian curve

FIG. 55. Final sum projection of acquisition 20 and its fitted gaussian curve
FIG. 56. Final sum projection of acquisition 21 and its fitted gaussian curve
FIG. 60. Modulation transfer function of acquisition 1
FIG. 61. Modulation transfer function of acquisition 2

FIG. 62. Modulation transfer function of acquisition 3
FIG. 63. Modulation transfer function of acquisition 4

FIG. 64. Modulation transfer function of acquisition 5
FIG. 65. Modulation transfer function of acquisition 6

FIG. 66. Modulation transfer function of acquisition 7
FIG. 67. Modulation transfer function of acquisition 8

FIG. 68. Modulation transfer function of acquisition 9
FIG. 69. Modulation transfer function of acquisition 10

FIG. 70. Modulation transfer function of acquisition 11
FIG. 71. Modulation transfer function of acquisition 12

FIG. 72. Modulation transfer function of acquisition 13
FIG. 73. Modulation transfer function of acquisition 14

FIG. 74. Modulation transfer function of acquisition 15
FIG. 75. Modulation transfer function of acquisition 16

FIG. 76. Modulation transfer function of acquisition 17
FIG. 77. Modulation transfer function of acquisition 18

FIG. 78. Modulation transfer function of acquisition 19
FIG. 79. Modulation transfer function of acquisition 20

FIG. 80. Modulation transfer function of acquisition 21
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BIOGRAPHICAL SKETCH

Rami R. Abu-Aita was born in Beit Jalla, Palestine, on November 22, 1979. Rami is the son of Rimon and Khitam Abu-Aita. Rami attended Bethlehem University, located in the city of Bethlehem, in 1997. After one year at Bethlehem University, he transferred to Francis Marion University in Florence, South Carolina. Rami graduated from Francis Marion University in December of 2001 with Bachelor of Science degrees in both computational physics and health physics. In January 2002, he enrolled in the medical physics program at the University of Florida and since then has been pursuing his Master of Science for which this thesis is a partial requirement.