Data-Driven Motion Detection and Characterization in PET Brain Scans Using List Mode

by

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Graduate Program in Medical Physics
Duke University

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James MacFall

Thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in the Graduate Program in Medical Physics in the Graduate School of Duke University

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ABSTRACT

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Abstract

Head motion during a Positron Emission Tomography (PET) brain scan can considerably degrade image quality. External motion-tracking devices have proven successful in minimizing this effect, but the associated time, maintenance, and workflow changes inhibit their widespread clinical use. List-mode PET acquisition allows for the retroactive analysis of coincidence events on any time scale throughout a scan, and therefore potentially offers a data-driven motion detection and characterization technique. An algorithm was developed to parse list-mode data, divide the full acquisition into short scan intervals, and calculate the line-of-response (LOR) midpoint average for each interval. These LOR midpoint averages, known as “radioactivity centroids,” were presumed to represent the center of the radioactivity distribution in the scanner, and it was thought that changes in this metric over time would correspond to intra-scan motion.

Several scans were taken of the 3D Hoffman brain phantom on a GE Discovery IQ PET/CT scanner to test the ability of the radioactivity to indicate intra-scan motion. Each scan incrementally surveyed motion in a different degree of freedom (2 translational and 2 rotational). The radioactivity centroids calculated from these scans correlated linearly to phantom positions/orientations. Centroid measurements over 1-second intervals performed on scans with ~1mCi of activity in the center of the field of
view had standard deviations of 0.026 cm in the $x$- and $y$-dimensions and 0.020 cm in the $z$-dimension, which demonstrates high precision and repeatability in this metric. Radioactivity centroids are thus shown to successfully represent discrete motions on the submillimeter scale. It is also shown that while the radioactivity centroid can precisely indicate the amount of motion during an acquisition, it fails to distinguish what type of motion occurred.
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1. Introduction

1.1 PET Brain Imaging

Positron Emission Tomography (PET) is an essential tool in neuroimaging. As a functional imaging technique, PET offers a glimpse at the way the body works. In brain imaging, the functional information provided by PET is of great value, particularly in disease diagnosis and staging, trauma evaluation, and treatment monitoring. Given the flexibility of PET with various radiotracers, the absolute quantification of its pixel values, and its interoperability with structural imaging techniques such as Computed Tomography (CT), PET has a unique role as a neuroimaging technique.¹

As a nuclear medicine modality, PET scanners detect radioactivity that has been injected into patients in the form of radionuclide-tagged chemical compounds, or radiotracers. Depending on the radiotracer used for the scan, the radioactivity can distribute in the body in various ways. Radiotracers are thus picked for each scan type to track specific physiology and accumulate in regions of interest. For PET brain imaging, radiotracers such as $^{18}$F-fluorodeoxyglucose (FDG) and $^{15}$O-labelled water have been commonly used to measure levels of neuronal activity in the brain, which can be useful in a large host of applications.² More recently, however, a steady influx of specialized radiotracers have kept PET on the cutting edge in the research and diagnosis of specific neurological diseases. One quintessential example is Pittsburgh Compound B, which highlights the signature beta-amyloid plaques of Alzheimer’s disease.³ With a variety of
radiotracers available, PET can be used to image different functionality in various parts of the body.

### 1.2 Principles of PET Imaging

PET scanning involves detecting the characteristic photon pairs produced by positron-emitting radiotracers. When a positron is emitted from a decaying nucleus, it travels a short distance (roughly 1 mm) before annihilating with an electron. This annihilation process results in two photons separated by nearly 180°, as shown in Figure 1.

![Figure 1: Positron annihilation with electron. When a positron (e+) is emitted from a decaying nucleus meets an electron (e-), the two annihilate, producing two photons (γ) separated by ~180°.](image)

The PET scanner itself comprises several concentric rings of square-faced scintillator crystals. These rings are designed to surround the patient so as to detect the photon pairs emanating from positron-electron annihilations, as shown in Figure 2.
Figure 2: PET scanner design. Subjects are surrounded by rings of square-faced scintillator crystal detectors. The blue arrows represent a photon pair emanating from a positron-electron annihilation. Each photon is detected by a scintillator crystal, shown in orange.

When two crystals simultaneously detect photons, a coincidence event is recorded. Coincidence events indicate that an annihilation occurred somewhere on the path between the two detectors, referred to as the line of response (LOR). Over the course of a scan, large numbers of coincidence events are recorded (~10^8 coincidences for a 6-min scan with 1 mCi of activity at the center of the scanner). After a scan is completed, various reconstruction algorithms can be used to generate an image from the coincidence events recorded.

*Two-dimensional (2D)* PET involves the use of septa between detector rings so as to only detect photon pairs within the same ring (or adjacent rings) of detectors, as shown in Figure 3a. This technique aids in the rejection of scatter events—coincidences involving photons whose trajectories have changed after deflecting off of electrons—and
random events—coincidence events recorded when two photons from different annihilations are detected simultaneously and wrongly associated with one another. The septa in 2D PET prevent out-of-plane photons from being detected, thereby reducing the number of image-degrading scatter and random events, at the expense of detection efficiency. In contrast, septa are not used in three-dimensional (3D) PET, as shown in Figure 3b, allowing for greater photon detection but enabling greater rates of scatter and random events.

![Diagram of 2D and 3D PET](image)

**Figure 3**: Cross sections of detector rings demonstrating 2D and 3D PET. In 2D PET (a), septa only allow the detection of photon pairs in the same or adjacent rings. In 3D PET (b), photon pairs can be detected in any two rings.

One of two methods can be used for recording PET coincidence data. Historically, the more commonly used method was called frame mode. In frame mode, each coincidence event contributes a count to its respective LOR bin in the raw data file “sinogram.” Though this method is more efficient in terms of storage space requirements, in recent years, the other method, list mode, has become increasingly more common. In list mode, coincidence events are written to the raw data file in sequence,
one after another. The list-mode raw data files also contain time markers recorded every millisecond during the scan (called time events). A schematic of list-mode data is shown in Figure 4. These time events effectively organize coincidence events by their detection times, giving the ability to generate images with coincidence data recorded at specific times. For example, if in a 6-minutelong brain scan, head motion is known to have occurred in the 5th minute, the list-mode file can be retrospectively “played back” to generate an image with only the first 4 minutes of coincidences.

<table>
<thead>
<tr>
<th>Coincidence Event 1</th>
<th>Coincidence Event 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>...</td>
</tr>
<tr>
<td>Coincidence Event 347</td>
<td></td>
</tr>
<tr>
<td>Time Event 1 (1ms)</td>
<td></td>
</tr>
<tr>
<td>Coincidence Event 348</td>
<td>Coincidence Event 349</td>
</tr>
<tr>
<td>...</td>
<td></td>
</tr>
<tr>
<td>Coincidence Event 691</td>
<td></td>
</tr>
<tr>
<td>Time Event 2 (2ms)</td>
<td></td>
</tr>
<tr>
<td>Coincidence Event 692</td>
<td></td>
</tr>
<tr>
<td>...</td>
<td></td>
</tr>
</tbody>
</table>

**Figure 4:** List-mode file structure. Coincidence events (green) are recorded in order, and time events (purple) are recorded every millisecond during acquisition.

In list-mode files, coincidence events are represented by the IDs of the two scintillator crystals hit. Each crystal has 2 IDs: the axial ID and the transaxial ID. The axial ID refers to the ring number of the detector (Figure 4b), whereas the transaxial ID refers to the in-ring position of the detector (Figure 4c). In the list-mode record, the two axial IDs and the two transaxial IDs are grouped, with the greater ID number referred to
as the “High” ID and the lesser ID number referred to as the “Low” ID. This convention is depicted in Figure 5.

1.3 Head Motion

One practical drawback of PET brain imaging is the likelihood of head motion during scans. Intra-scan head motion can significantly reduce image quality and quantitative accuracy. When motion occurs, the detector crystal pairs associated with each subsequent coincidence event change, ultimately causing a blurring effect. Such image degradation occurs with even relatively minor motion; according to Menke et al., “[i]n a cylindrical scanner with 380 mm diameter and a crystal size of approximately $6.5^2$ mm$^2$ … a head rotation of less than one degree can change the discrete coordinates of a
central line of response.”

Figure 6 illustrates this point. In general, the degree of image degradation depends on both the fraction of coincidence events recorded after a motion occurs and the magnitude of the motion.

![Image of Figure 6: Effects of intra-scan head motion. Less than 1° of rotation causes the coordinates of a central line of response (blue) to change (orange).](image)

The blurring caused by intra-scan motion reduces the effective resolution of the image and skews the pixel values. Furthermore, motion leads to misalignment of the PET and CT images, leading to inaccurate attenuation correction. Figure 7 shows a pertinent example of a PET brain scan in which head motion occurred during acquisition. List-mode data was used to reconstruct the image both before and after the onset of motion.
Figure 7: PET brain scan before (left) and after (right) the onset of head motion.
Images courtesy of Timothy Turkington, Ph.D.

As a rigid body, the head has 6 degrees of freedom: 3 translations (surge, sway, and heave), and 3 rotations (roll, pitch, and yaw). Surge is translation along the z-axis, sway is along the x-axis, and heave is along the y-axis. These translations are shown in Figure 8. Roll is rotation about the z-axis, pitch is about the x-axis, and yaw is about the y-axis. These rotations are shown in Figure 9.

Figure 8: Head translations. Surge occurs in the z-dimension (axially), sway in the x-dimension (laterally), and heave in the y-dimension (anterior-posterior).
Figure 9: Head rotation. Roll occurs about the z-axis, pitch about the x-axis, and yaw about the y-axis.

While head motion is commonly exhibited in patients with neurodegenerative disorders such as Parkinson’s Disease, it is not limited to just these cases. In a 1994 study, Green et al. found that even on healthy patients fitted with thermoplastic head restraints, detectable levels of head motion in PET brain scans still pervade. In order to ensure optimal image quality and reliable quantitation, the need for motion correction techniques in PET brain imaging becomes apparent.

### 1.4 Existing Solutions

Several methods for reducing the unwanted effects of head motion during PET brain scans have been explored and indeed successfully developed. In general, these include three main steps:

1. *Motion detection*, establishing if and when motion occurs in a scan

2. *Motion characterization*, determining the type and amount of motion
3. *Motion correction*, reversing the adverse effects of motion by manipulating either the PET raw data or the reconstructed image.

Techniques for the last step, motion correction, are generally well-established and can be used in conjunction with a variety of techniques for the first two steps, motion detection and characterization. As such, existing motion correction techniques will be presented first, followed by motion detection and characterization techniques.

### 1.4.1 Motion Correction Techniques

One method of motion correction in PET scans is the multiple acquisition frame (MAF) motion correction technique described by Picard and Thompson.\(^8\) Frame-mode acquisition is used in conjunction with a motion detection system in such a way that when motion is detected, a new acquisition frame is triggered, in effect creating a distinct raw data file for each discrete head position during the scan. The MAFs produced during the scan are separately reconstructed, after which the resulting images are appropriately transformed and summed together. This method is ideal for PET systems that do not offer list-mode acquisition.

Another motion correction method is LOR rebinning, proposed by both Menke et al.\(^6\) and Daube-Witherspoon et al.\(^9\) In this technique, individual LORs are changed to reflect the pre-motion position of the patient. While this method offers individual correction of every event, it requires list-mode acquisition or a specialized interface to allow for real-time LOR correction.
1.4.2 Motion Detection and Characterization

1.4.2.1 External Motion-Tracking Devices

One approach to motion detection and characterization involves using external motion-tracking devices in sync with PET acquisition. Using the MAF motion correction technique, in 2002 Fulton et al. described using the Polaris optical motion-tracking system (Northern Digital Inc., Waterloo, ON, Canada) to trigger a new acquisition frame at the onset of head motion.\(^10\) This effort largely yielded successful results, with only minor deviations between motion-free control images and the motion-corrected images. A 2003 corollary study by Woo et al. hinged on the same motion-tracking system for motion detection and correction, this time with list-mode and the LOR rebinning technique.\(^11\) While this technique successfully reduced motion artifacts, the motion-corrected images from the MAF method would prove “smoother.”

1.4.2.2 Data-Driven Motion Tracking

Though successful, the aforementioned motion detection and characterization methods still have not seen widespread clinical use. This is likely due to the extra time, maintenance, and workflow changes associated with external motion-tracking devices. More practical motion detection and correction methods in PET would involve little or no extra work for technologists and work in a variety of workflows and clinical contexts.

Alternative approaches to the use of motion-tracking devices involve analyzing raw PET data for signs of motion—“data-driven” approaches. In a 2001 manuscript,
Klein et al. describe a data-driven technique in which the list mode data stream is used to compute the sinogram “center of mass” over time. The metric referred to as “center of mass” is really an average of axial planes weighted by the number of counts in each—in essence an axial centroid of coincidence events. In this study, changes in this centroid metric indicated motion in the axial dimension. Though this method was developed for tracking motion in cardiac PET, the concepts introduced in this work can be extended to motion detection in brain scans.

1.5 This Work

The impetus for this work was to develop an easily implementable motion detection and characterization technique for PET brain scans, to be potentially used with one of the previously established motion correction techniques. To avoid the complexity of external motion tracking devices, we sought to use the list-mode data stream for a data-driven technique, as in the work of Klein et al. Our aim was to use simple and fast count-based statistics in list-mode data to find signatures of motion in PET brain scans.

In this work, we present the LOR-midpoint average, referred to as the “radioactivity centroid,” as an indicator of intra-scan motion. This metric was presumed to represent the center of the radioactivity distribution in the scanner, and it was thought that changes in this metric over time would correspond to intra-scan motion. In the remainder of this work we assess the radioactivity centroid as an indicator of motion.
2. Methods

2.1 Algorithm

An algorithm was written in the C programming language to read in list-mode data and calculate the radioactivity centroid over time. The user chooses the file increment size (the number of seconds of coincidences to process at a time). For each increment, the algorithm finds the LOR midpoints and averages them. The algorithm then prints out the spatial coordinates of the radioactivity centroid for each time increment. Figure 10 is a flow chart that demonstrates this process.

Figure 10: Flow chart of radioactivity centroid calculation. Inputs to the algorithm are the list-mode file and the user-chosen file increment size. For each increment, the LOR midpoints are calculated and averaged. The output is the centroid coordinates for each file increment.

The details of the LOR midpoint determination and averaging are covered further in the subsequent sections.
2.1.1 LOR Midpoint Coordinates

The algorithm determines radioactivity centroid by first treating the geometric centers of each LOR as the point of positron annihilation. Because of the cylindrical scanner geometry, these points are determined in cylindrical coordinates relative to the center of the scanner, as shown in Figure 11.

![Scanner coordinate system](image)

**Figure 11**: Scanner coordinate system. Origin $O$ is located at the geometric center of the scanner. Point $P$ is represented by coordinates $r$, $\theta$, and $z$: respectively, the distance from the $z$-axis, the polar angle, and the axial distance from the origin.

2.1.1.1 Axial

To find the axial component of the midpoint of each LOR, the algorithm utilizes the same method as is used in a seminal technique designed for simplifying the reconstruction of 3D PET data known as “single-slice rebinning.” In this technique, a coincidence from a cross-slice (i.e. a coincidence with differing high and low axial IDs) is
treated as having occurred at the axial position midway between its high and low axial IDs. The high and low axial IDs are thus reassigned to the average axial ID. This was a necessary approximation during the dawn of 3D PET when exact treatment of each photon trajectory was not computationally feasible.

The axial component of the centers of each LOR—represented by the cylindrical coordinate $z$—is assumed to be the slice midway between the two axial detector IDs. For each coincidence event, the ID numbers of the top and bottom detectors are averaged, effectively attributing the midway axial position to the event. As illustrated in Figure 12, this approximation is more precise for centrally-located annihilations than for peripherally-located annihilations.

![Diagram showing determination of $z$.](image)

Figure 12: Determination of $z$. For the LOR shown in blue, top ring ID “H” and bottom ring ID “L” are averaged to approximate the axial position of the annihilation.
The different possible annihilations (red circles) show how the approximation becomes less precise as radial position increases.

2.1.1.2 Transaxial

To determine the in-plane transaxial location of each coincidence event, transaxial detector IDs are first mapped to their respective polar angles. The first step is to reverse the ordering of the transaxial IDs. Transaxial IDs increase in a clockwise fashion with respect to the front of the gantry, whereas polar angles conventionally increase counterclockwise about the origin. To reverse the order, the transaxial IDs of each coincidence event are subtracted from \( N \), the total number of transaxial crystals. To map the ID range from 0 to \( N-1 \) (instead of from 1 to \( N \)), a modulo \( N \) operation is subsequently performed. This is demonstrated in Figure 13 for the simplified \( N = 8 \) case.

![Figure 13](image)

**Figure 13:** Transaxial ID reversal, demonstrated on a simplified, 8-detector ring. To match counterclockwise polar angle convention, ID number ordering is reversed.

The next step in the algorithm is to rotate the newly reversed transaxial IDs such that \( id\#_0 \) is at the 0° position. This is accomplished by adding the detector displacement, \( d \), to each detector ID, modulo \( N \). This step is demonstrated in Figure 14, with \( N = 8 \) and \( d = 3 \).
Figure 14: Transaxial ID alignment with 0°, demonstrated on a simplified, 8-detector ring with displacement of 3. To align the 0 ID with 0°, 3 is added to each ID mod 8.

The final step in mapping transaxial IDs to respective polar angles is to multiply each freshly rotated detector ID by the detector-angle conversion factor $360° / N$. This step is shown in the 8-detector setup in Figure 15.

Figure 15: Transaxial ID-to-angle conversion, demonstrated on a simplified, 8-detector ring. Each ID is multiplied by the number of degrees in a full rotation divided by the number of crystals in a full rotation, or $360° / 8$.

The full transaxial-ID-to-polar-angle formula is given by:

\[
\theta_n = \frac{360°}{N} \cdot [(N - n + d) \mod N] \tag{1}
\]
where $n$ is the transaxial crystal ID, $\theta_n$ is the polar angle for crystal $n$, $N$ is the number of transaxial crystals in the ring, and $d$ is the offset. The combined effect of these steps is shown in Figure 16 on a 72-detector ring.

![Figure 16: Polar angle mapping. Transaxial crystal IDs are assigned appropriate angles above horizontal.](image)

After each transaxial detector ID is mapped to its respective angle, the polar coordinate $\theta$ of the center of the LOR is determined. First, the difference between the high transaxial crystal angle $\theta_2$ and the low transaxial crystal angle $\theta_1$ is found. When this difference is less than or equal to 180°, the two angles are simply averaged to determine $\theta$, as shown in Figure 17a. The formula for $\theta$ in this case is:

$$\theta = \frac{\theta_1 + \theta_2}{2} \quad (2)$$

When the difference between $\theta_2$ and $\theta_1$ is greater than 180°, averaging yields an angle that is 180° too large, as shown in Figure 17b. The formula for $\theta$ is thus corrected in the following way:
\[ \theta = \frac{\theta_1 + \theta_2}{2} - 180^\circ \]  \hspace{1cm} (3)

Figure 17: Determination of \( \theta \). In (a), the angular difference between the high and low crystals is \(< 180^\circ\), so the average of the two angles is taken. In (b), the angular difference is \(> 180^\circ\), so a 180° term is subtracted from the average to avoid generating the incorrect angle, shown in purple.

Next, the radial component \( r \) of the LOR center is determined. Geometrically, the transaxial projection of the LOR is a chord, meaning that the line segment through the scanner center that bisects the LOR is also perpendicular to it. A right triangle can be formed with the LOR, the perpendicular bisector of LOR, and a radius to the low transaxial detector. This geometry is highlighted in Figure 18. With this information, \( r \) is given by:

\[ r = R \cos \alpha \]  \hspace{1cm} (4)

where \( R \) is the radius of the scanner (center-to-crystal distance) in cm, and \( \alpha \) is

\[ \frac{\theta_2 - \theta_1}{2} \text{ when } \theta_2 - \theta_1 \leq 180^\circ \text{ or } \frac{\theta_2 - \theta_1}{2} - 180^\circ \text{ when } \theta_2 - \theta_1 > 180^\circ. \]
Figure 18: Determination of $r$. The LOR (blue) is perpendicularly bisected by $OP$ (red), where $O$ is located at the center of the ring. A right triangle is formed by the LOR, $OP$, and the line segment connecting $O$ and crystal $\theta_1$. Cylindrical coordinate $r$ is one leg of the triangle, and can be determined trigonometrically.

The midpoint approximation has the cumulative effect of “underselling” the true radial position of a radioactive point in space. As demonstrated in Figure 19, the midpoints of all possible LORs from a point source yield a centroid with approximately half the radial coordinate of the point source. To counteract this radial underselling, the $r$ coordinate of each midpoint is multiplied by a factor of 2.
Figure 19: “Radial underselling.” For a radioactive point source (black point) displaced from the scanner center (blue point), the centroid of all possible LOR midpoints (red points) is approximately half of the true value (purple “X”).

2.1.2 Sensitivity Weighting

The probability of coincidence detection depends on the position of the annihilation within the scanner. When computing the radioactivity centroid, each LOR midpoint needs to be weighted according to the probability of detection at that point, otherwise the centroid measurement will lean towards the regions with higher detection probability. After the coordinates of the LOR midpoint are determined, the algorithm computes the weighting factors associated with this point.

Before addressing the weighting methods used in the algorithm, we establish pertinent semantics. As a PET performance metric, “sensitivity” commonly refers to the ratio of counts detected per second to source activity. This is usually a single number.
associated with a particular scanner. Here, “sensitivity” refers to the relative likelihood of coincidence detection based on annihilation position in the scanner. Each LOR midpoint is attributed an axial sensitivity and a transaxial sensitivity.

2.1.2.1 Axial Sensitivity

In 3D PET, which was used in this work, axial sensitivity exhibits symmetry about the $z = 0$ plane, increasing linearly as axial position approaches 0 (scanner center) in both directions. The axial sensitivity graph is depicted for a 15-plane PET system in Figure 20.

![Figure 20: 3D PET axial sensitivity.](image)

The axial weight of each LOR midpoint is the inverse of its axial sensitivity. For axial position $z$ (in terms of plane number) in a scanner with $P$ axial planes, the axial weight $w_z$ is given by:
\[ w_z = \begin{cases} \frac{1}{2z + 1}, & \text{for } z \leq \frac{p + 1}{2} \\ \frac{1}{p + 1 - (2z + 1)}, & \text{for } z > \frac{p + 1}{2} \end{cases} \] (5)

2.1.2.2 Transaxial Sensitivity

Unlike axial sensitivity, in-plane transaxial sensitivity is a function of two variables—\( r \) and \( \theta \)—and it varies in form among PET scanners. To find the transaxial sensitivity for a PET scanner, count rates need to be collected with a positron-emitting line source at various positions to produce count rate as a function of \( r \) and \( \theta \), as illustrated in Figure 21. For the purposes of this work, transaxial sensitivity was assumed to just be a function of radial displacement (constant with respect to polar angle). This procedure is discussed further in 2.6.1, as this sensitivity test was performed on the PET scanner used for experimentation.

Figure 21: Transaxial sensitivity testing. A line source is placed at varying radial positions, and count rates are recorded at each position.
The results of the transaxial sensitivity test performed are presented in 3.1, with a formula for sensitivity as a function of \( r \) given in equation 10. With these results, the transaxial weight \( w_r \) is determined by inverting the sensitivity function in equation 10:

\[
\frac{1}{2000 \text{sinc}(2r + 0.55) + 200r^2 + 1000r + 700000}
\]

(6)

2.1.3 Centroid Calculation

After the weights are determined for each LOR midpoint, the centroid is calculated. The \( x \), \( y \), and \( z \) (Cartesian) coordinates of the midpoints are each averaged separately. For \( L \) midpoints, the following formulae are used:

\[
x_c = \frac{\sum_{i=1}^{L} w_{r,i} x_i}{\sum_{i=1}^{L} w_{r,i}}
\]

(7)

\[
y_c = \frac{\sum_{i=1}^{L} w_{r,i} y_i}{\sum_{i=1}^{L} w_{r,i}}
\]

(8)

\[
z_c = \frac{\sum_{i=1}^{L} w_{z,i} z_i}{\sum_{i=1}^{L} w_{z,i}}
\]

(9)

The centroid position of these midpoints is thus given by \((x_c, y_c, z_c)\).

2.2 Phantoms

Three phantoms were used in scans to test the aforementioned algorithm. One phantom was the 3D Hoffman brain phantom,\(^{14}\) displayed in Figure 22, developed by Data Spectrum Corporation (Durham, NC). The Hoffman phantom is a fillable cylinder—17.5 cm in height and 20.8 cm in diameter—with plastic slab inserts cut to resemble axial slices of the human brain. Differing thicknesses within each slab create the apparent 4:1
uptake ratio between gray and white matter. The other two phantoms used were a uniform cylindrical phantom and a 40.2-μCi $^{68}$Ge line source.

![Figure 22: 3D Hoffman brain phantom.](image)

### 2.3 Stages

Scans were performed to survey surge, sway, roll, and yaw motions of the Hoffman brain phantom (see Figures 8 and 9 for reference on head motions). Three of these scans utilized specially made stages. The stages were designed to hold the Hoffman brain phantom in the scanner, while allowing for controlled, precise motions in each degree of freedom studied. Two of the stages used a foam cradle. The foam cradle, pictured in Figure 23, was made from stiff foam that would not deform under the weight of the water-filled phantom. It was mounted to the sway and yaw stages for their respective scans.
Figure 23: Foam cradle.

The foam cradle was used for the sway stage and the yaw stage. The sway stage, designed to only allow for lateral motion, used two pairs of plastic tracks. One pair was fastened to the underside of the foam cradle, while the other was fastened to a sheet of poster board that sat atop the scanner bed. The friction of the tracks enabled small, controlled motions of <5 mm. The poster board was marked with 5 mm increments in the lateral dimension. The sway stage is pictured in Figure 24.
Figure 24: Sway stage.

The yaw stage also used the foam cradle. The yaw stage was designed to only allow for rotational motion about the $y$-axis. To accomplish this, a microwave roller was placed atop a poster board labelled with angular increments. The foam cradle sat atop the microwave roller, with a plastic screw securing the center of the cradle to the center of rotation on the poster board. The Yaw stage is shown in Figure 25.
Figure 25: Left: Microwave roller on labelled poster board. Right: Yaw stage with foam cradle.

The final stage, the roll stage, was designed to only allow for motion about the z-axis. This was accomplished using two parallel 1.25”-diameter PVC pipes. Two 2x4’s were used to secure the pipes; each end of the pipes was fed into a hole drilled into a 2x4. These parallel pipes held the Hoffman phantom and allowed the user to manually and precisely rotate it about the z-axis without moving it in other degrees of freedom. The stage is shown in Figure 26.

Figure 26: Roll stage.
2.4 Radioactive Sources

For their respective scans, the Hoffman and uniform cylindrical phantoms were filled with 0.8-1.3 mCi of aqueous FDG. Another scan used a 40.2-µCi $^{68}$Ge line source to gauge scanner sensitivity.

2.5 Scanner

The PET/CT scanner used for this project was the 5-ring version of the GE Discovery IQ (GE Healthcare, Milwaukee, WI). The scanner 5 parallel rings of 288 detector blocks. Each block has an 8x8 array 6.5-mm x 6.5-mm bismuth germanate (BGO) scintillator crystals (not capable of time-of-flight scanning). The Discovery IQ is pictured in Figure 27.

![Image](image.png)

Figure 27: Left: The GE Discovery IQ, with the Hoffman brain phantom. Right: GE Discovery IQ block detector arrangement. Each square face contains an 8x8 array of BGO scintillator crystals.

2.6 Data Acquisition

Seven scans were performed to calibrate and test the algorithm. Two, Radial Sensitivity and Constancy, were used for error-checking and minor adjustments. Four
others—Surge, Sway, Roll, and Yaw—were used to determine the accuracy of the centroid calculation in different dimensions. A final scan was performed to test the algorithm in characterizing a single small, arbitrary motion during a mock brain scan.

### 2.6.1 Radial Sensitivity

The Radial Sensitivity scan was performed to determine the radial weights of the transaxial components of the LOR midpoints, as discussed in 2.1.2.2. The $^{68}$Ge line source, fixed in a holder that allowed for controlled motion in the $x$-dimension, was positioned in the scanner parallel to the $z$-axis. The holder and bed height were adjusted to position the source in the transaxial location with the highest count rate, as reported by the PET scanner. This location was presumed to be the $r = 0$ point in the scanner. The source was then translated laterally: first in steps of 2.0 mm, near the center, then in steps of 5.0 mm, and finally in steps of 10 mm. A 10-s acquisition was taken at each position. The list mode files for the scans were fed into the algorithm with time increments of 10 s. The line source and holder are pictured in Figure 28.

![Figure 28: $^{68}$Ge line source and holder in scanner for radial sensitivity scan.](image-url)
2.6.2 Constancy

The Constancy scan was performed with the uniform cylinder to demonstrate the centroid measurements remained constant over an extended period of time when no motion was present, and to explore the effects of time-increment length on algorithm precision. The uniform cylindrical phantom remained stationary in the center of the PET scanner over a 10-min acquisition. The list-mode file was run through the algorithm in time increments of 1 s and 2 s.

2.6.3 Translations

Two scans were performed to test the accuracy of the centroid in representing intra-scan translational motion. The two scans consisted of Surge and Sway, surveying $z$-axis and $x$-axis translation, respectively. Due to the radial symmetry of the PET scanner, the centroid behavior was assumed to be the same for motion along the $x$- and $y$-axes. Therefore, the Sway scan alone would be sufficient for surveying both sway and heave motions.

For the Surge scan, the scanner bed was used to translate the Hoffman brain phantom in 10-mm steps along the entire $z$-axis of the PET scanner. A 30-s scan was taken at each position. For the Sway scan, the sway stage was used to move the Hoffman phantom laterally in 5-mm increments. The phantom was scanned for 15s at each position. Both scans were run through the algorithm with 1-s time increments. An image taken during the Sway scan is shown in Figure 29.
2.6.4 Rotations

Two additional scans were performed to test the accuracy of the centroid in representing intra-scan rotational motion. The two scans were Roll and Yaw, surveying rotation about the z-axis and y-axis, respectively. As with the Surge scan, the Yaw scan alone was assumed to be sufficient for testing centroid behavior for both yaw and pitch rotations.

The roll and yaw stages were used in their respective scans to rotate the phantom in 5° steps over a 360° trajectory. The phantom was scanned for 30 s in each orientation. Both scans were run through the algorithm with 1-s increments. The Roll and Yaw scans are depicted in Figure 30.
2.6.5 Arbitrary Motion

The Arbitrary motion scan was performed to test the algorithm in the detection and characterization of a small, arbitrary motion. The Hoffman brain phantom was placed in the PET scanner for a 6-minute acquisition. At one point during the scan, the phantom was moved slightly, only once, in a way that combined multiple degrees of freedom. The list mode file was run through the algorithm in 1-s increments.
3. Results

3.1 Radial Sensitivity

The radial sensitivity of the GE Discovery IQ as measured in the Radial Sensitivity scan is represented as coincidence counts detected during the 10-s increments vs. lateral position. The plot was fit with the following function:

\[ S(r) = 2000 \text{sinc}(2r + 0.55) + 200r^2 + 1000r + 700000. \]  

The radial sensitivity data along with the fit are shown in Figure 31.

![Figure 31: Radial sensitivity of the GE Discovery IQ.](image)

The inverse of the fit function was used for the transaxial weighting function, \( w_r \), in the centroid calculation (Equation 6).

The 10-s centroid calculations from the Radial Sensitivity scan using these weights are shown against the true phantom position in Figure 32 (x-dimension only). The R\(^2\) value for the true x-positions of the line source and the centroid x-coordinates is
0.998. The largest outlier occurs at \( x = -19.6 \text{ cm} \), where the centroid is calculated to be -20.65 cm, an error of 1.05 cm.

**Figure 32:** Centroid calculation \( x \)-coordinate vs. true \( x \)-position. The \( R^2 \) value is 0.998. The largest outlier occurs at \( x = -19.6 \text{ cm} \), where the centroid is off by 1.05 cm.
3.2 Constancy

The 1-s centroids from the Constancy scan are plotted over time in Figure 33. The 2-s centroids are plotted over time in Figure 34.

Figure 33: 1-s Constancy scan centroid measurements over time.

Figure 34: 2-s Constancy scan centroid measurements over time.

The averages and standard deviations for each coordinate of the centroid measurements are shown in Table 1.
Table 1: Statistics for 1-s and 2-s Constancy scan centroids.

<table>
<thead>
<tr>
<th></th>
<th>1 s</th>
<th>2 s</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Average (cm)</td>
<td>Standard Deviation (cm)</td>
</tr>
<tr>
<td>x</td>
<td>0.931</td>
<td>0.026</td>
</tr>
<tr>
<td>y</td>
<td>5.861</td>
<td>0.026</td>
</tr>
<tr>
<td>z</td>
<td>-2.650</td>
<td>0.020</td>
</tr>
</tbody>
</table>

3.3 Translations

For the two Translation scans, the 1-s centroid measurements are plotted against the corresponding phantom position in the dimension with intended motion to show the effectiveness of the centroid as an indicator of position. The standard deviations of the centroids at each position are also shown to indicate reliability and precision of the centroid metric. Finally, the 1-s centroids in the dimensions intended to remain stationary are shown against time to demonstrate any motion contamination—or unintended/unexpected motion in other dimensions.
3.3.1 Surge

The centroid measurements for the Surge scan are plotted against Hoffman brain phantom axial displacement in Figure 35. Each phantom position has a cluster of 30 1-s centroid measurements.

![Graph showing Surgescan centroids vs phantom position, z-coordinate.](image)

Figure 35: 1-s Surge scan centroids vs phantom position, z-coordinate. The figure on the right shows the inner 20 cm, where linear centroid behavior is observed.

Standard deviations of the 1-s centroid measurements at each position are displayed in Figure 36.
Figure 36: Standard deviations of 1-s Surge scan centroids vs. phantom position, z-coordinate.

1-s Surge scan centroid measurements in the x- and y-dimensions are shown in Figure 37.

Figure 37: 1-s Surge scan centroids vs. time, x- and y-coordinates. The red rectangle indicates when the phantom was fully in the field of view.
3.3.2 Sway

The centroid measurements for the Sway scan are plotted against phantom position in Figure 38. Each phantom position has a cluster of 15 1-s centroid measurements.

![Graph showing the relationship between centroid and phantom position](image)

**Figure 38: 1-s Sway scan centroids vs phantom position, x-coordinate. The figure on the right shows the inner 2 cm.**

Standard deviations of the 1-s centroid measurements at each position are displayed in Figure 39.
Figure 39: Standard deviations of 1-s Sway scan centroids vs. phantom positions, x-coordinate.

1-s Sway scan centroid measurements in the z- and y-dimensions are shown in Figure 40.

Figure 40: 1-s Sway scan centroids vs. time, z- and y-coordinates.
3.4 Rotations

For the two Rotation scans, the 2 dimensions of the 1-s centroid measurements exhibiting motion are plotted against one another to show the effectiveness of the centroid in representing the phantom orientation. For the same purpose, centroid angles are shown against the corresponding rotation angles of the phantom. The standard deviations of the centroid angles at each orientation are shown to indicate reliability and precision of the centroids. Finally, the 1-s centroids in the dimensions intended to remain stationary are shown against time to demonstrate any motion contamination.

3.4.1 Roll

Results for the Roll scan are shown with the x- and y-coordinates of the centroid as well as the polar angle of the centroid in the xy-plane. The x- and y-coordinates of the 1-s centroid measurements for the roll scan are shown in Figure 41.
The $xy$-polar angles of the centroid measurements were calculated as $\tan^{-1}\frac{y_c}{x_c}$.

From Figure 38, the center of rotation was determined to be located at (-0.35 cm, 1.15 cm). Both the polar angles about the origin and the polar angles about the center of rotation (COR) are shown in Figure 42, along with the true phantom orientation angle in the $xy$-plane.
Figure 42: 1-s Roll scan centroids vs. phantom orientation, \( xy \)-polar angle.

Standard deviations of the 1-s centroid measurements at each orientation are displayed in Figure 43.

Figure 43: Standard deviations of 1-s Roll scan centroids vs. phantom orientations, \( xy \)-polar angle.
Centroid measurements in the $z$-dimension are shown in Figure 44.

![Graph showing 1-s Roll scan centroids vs. time, $z$-coordinate.](image)

**Figure 44: 1-s Roll scan centroids vs. time, $z$-coordinate.**

### 3.4.2 Yaw

Results for the Yaw scan are shown with the $x$- and $z$-coordinates of the centroid as well as the polar angle of the centroid in the $xz$-plane. The $x$- and $z$-coordinates of the 1-s centroid measurements for the Yaw scan are shown in Figure 45.
Figure 45: 1-s Yaw scan centroids, \( z \)-coordinates vs. \( x \)-coordinates

The \( xz \)-polar angles of the centroid measurements were calculated as \( \tan^{-1} \frac{z_c}{x_c} \).

Using Figure 43, the center of rotation was determined to be located at the origin, \((0 \text{ cm, } 0 \text{ cm})\). The polar angles are shown in Figure 46, along with the true phantom orientation angle in the \( xz \)-plane.
Figure 46: 1-s Yaw scan centroids vs. phantom orientation, $xz$-polar angle.

Standard deviations of the 1-s centroid measurements at each orientation are displayed in Figure 47.

Figure 47: Standard deviations of 1-s centroids vs. phantom orientations, $xz$-polar angle.
Centroid measurements in the $y$-dimension are shown in Figure 48.

![Figure 48: 1-s centroids vs. time, $y$-coordinates.](image)

### 3.5 Arbitrary Motion

The list-mode file for the Arbitrary Motion scan was run through the algorithm to check for motion. The plots in Figure 49 show each centroid coordinate vs. time.

![Figure 49: 1-s Arbitrary Motion scan centroids vs. time.](image)
These plots indicate a discrete motion having occurred 141 s into the scan. Using the 141-s mark as the breaking point, the image was reconstructed using just the pre-motion coincidence data then just the post-motion coincidence data. Five axial slices are shown for the original, pre-motion, and post-motion image sets in Figure 50.

Figure 50: Comparison of original, pre-motion, and post-motion image sets from the Arbitrary Motion scan. Top: original. Middle: pre-motion. Bottom: post-motion. All images reconstructed with OSEM iterative reconstruction, 8 iterations, 12 subsets.
4. Discussion

4.1 Algorithm Performance

4.1.1 Centroid Behavior

The results of the various scans demonstrate the strength of the radioactivity centroid as an indicator of intra-scan motion. The Constancy scan showed that for a subject positioned in the middle of the scanner with approximately 1mCi of activity, the $x$- and $y$-coordinates of 1-s centroids have uncertainties of 0.026 cm, and the $z$-coordinates have an uncertainty of 0.020 cm, as shown in Table 1. Predictably, when increment length is doubled, the number of coincidences in each increment doubles and these uncertainty figures decrease by a factor of $\sqrt{2}$.

The two Translation scans shed further light on the behavior of the centroid produced by the algorithm. Figures 35 and 38 reveal that the centroid follows phantom position linearly. While in the Surge scan the linearity seemingly breaks down outside of the inner 20-cm in the axial dimension, this can be explained by the Hoffman phantom beginning to exit the PET scanner at these positions. The Surge scan shows that the uncertainty in the $z$-coordinate of the centroid increases with axial displacement, as shown in Figure 36. This is likely due to the increased random-to-true-events ratio near the edge of the FOV. As the phantom leaves the PET scanner, the rate of true events decreases faster than does the rate of random events, decreasing the centroid precision. Regardless, the greatest standard deviation recorded was 0.08 cm. This occurred when
the majority of the phantom was outside of the axial field of view (FOV). In the Sway scan, any correlation between lateral position and standard deviation was unnoticeable, with most values falling between 0.01 and 0.03 cm, as shown in Figure 39.

The two Rotational scans demonstrated the centroid behavior for motion in multiple dimensions. The Roll scan involved motion in x and y, and the Yaw scan involved motion in x and z. For each scan, the plots of the two moving centroid coordinates in Figures 41 and 45 show the rotational trajectories of the phantom as well as its centers of rotation (the plot for the Roll scan indicates that the phantom was not properly centered at the origin). The outliers in the Roll scan are presumed to be a consequence of the centroid falling near or on the origin; as \( r \) decreases, the \( \theta \) measurement becomes less precise. The asymmetric trajectory for the Yaw scan is likely due to the increased self-attenuation of the phantom at certain orientations. The centroid polar angles followed the true phantom rotation angles linearly, as shown in Figures 42 and 45, albeit with different uncertainties. The Yaw scan proved considerably more precise, with standard deviations primarily falling between 0.2° and 0.6°, as shown in Figure 47. In contrast, the standard deviations of the Roll scan fell between 5° and 10°, depicted in Figure 43.

The results of the Random Motion scan indicate the ability to precisely determine the time of intra-scan motion using centroid measurements. With a noticeable reduction
in blurring, the pre- and post-motion images confirm that motion indeed occurred 141 s into the scan, as shown in Figure 50.

### 4.1.2 Motion Contamination

Both Translation scans were subject to motion contamination; Figures 37 and 40 show centroid changes in dimensions that were intended to be motion-free. For the Surge scan, the $x$- and $y$-coordinates of the centroids gradually travel roughly 0.5 cm from their starting positions, while for the Sway scan this travel amounted to roughly 1 cm. The cause of this motion contamination is unknown. Potential causes include imprecise phantom and stage positioning, centroid skewing due to increased rates of random events near the edge of the FOV, and improper sensitivity weighting near the edge of the FOV. Future scans will be performed to further explore the causes of this motion contamination.

Figures 44 and 48 indicate that motion contamination was minimal in the Rotation scans.

### 4.2 Experimental Limitations

One sizable experimental limitation was that the polar component of transaxial sensitivity of the scanner was assumed to be constant (first mentioned in 2.1.2.2). This is not a trivial assumption, particularly for the scanner used in the experiments, which exhibited relatively dynamic changes in count rate with changing transaxial position. While this assumption made for a simpler sensitivity experiment, a better model of
scanner sensitivity could yield more precision in centroid measurements, particularly at the edge of the FOV.

Another experimental limitation was the aforementioned motion contamination observed in the two translational scans. Despite efforts to solely induce motion in the dimension studied, centroid measurements indicated motion in other dimensions. It is unclear from the results whether this was due to an experimental flaw or a phenomenon in the centroid measurement that requires consideration. One possible explanation is the increased random-to-true-events ratio near the edge of the FOV. Whether or not this was the cause of motion contamination, it deserves consideration. Whatever the cause, more scans need to be performed to better understand the motion contamination observed.

Some of the sources of error in the centroid measurement could have been better understood with CT images. CT images of the Hoffman phantom at its various positions and orientations would serve as a good reference—a “ground truth” of sorts—when assessing centroid accuracy. Using the CT would have helped determine the causes of motion contamination and decreased precision at the edges of the FOV.

A further experimental weakness was the lack of continuous motion in the scans. The scans all surveyed discrete motions, which were followed by stationary acquisition periods. Intra-scan head motions are not necessarily discrete; the motions themselves can last for non-trivial periods of time. To further assess the effectiveness of the centroid technique, continuous motion needs to be considered and tested.
One potential issue when using the radioactivity centroid clinically would be the effects of out-of-field activity. In a patient, radiotracers such as FDG are distributed throughout the body. The out-of-field activity could potentially contribute random events, which could decrease the precision and accuracy of the centroid measurements. This was not considered in this work. Future work needs to focus on the effects of out-of-field activity.

4.3 Pitfalls

One major pitfall of the centroid technique is its inability to distinguish between rotational and translational motions of the phantom. Individual changes in the centroid show only the positional changes of the center of radioactivity; it does not indicate how the subject moved to induce that centroid change. A change in centroid measurement could be due to a translational or rotational motion of the subject, as depicted in Figure 51. While the centroid shows promise for characterizing the amount of a certain motion, it fails to sufficiently show what motion occurred. As such, the technique succeeds in motion detection, but not entirely in motion characterization.
Figure 51: A centroid change can be due to either a translation (left), a rotation (right), or a combination of both.

Another noteworthy weakness of the centroid technique is its dependence on separation between the centroid and the COR, in cases of rotational motion. If the centroid falls on the COR, the rotation goes undetected. This phenomenon is illustrated in Figure 52.

Figure 52: For rotational motion, the centroid technique requires separation between the centroids (X’s) and the COR to detect motion.
The Roll scan exhibited near alignment of the centroid and the COR—the centroid measurements were between 1- and 2-mm distance from the COR. This could explain the conspicuously high standard deviations of the Roll scan compared to those of other motions. In general, due to the right-left radial symmetry of the brain, centroid measurements may fall close to the roll COR, meaning that roll may be less precisely detected.

4.4 Future Developments

One major future development for the centroid technique is the implementation of a standard by which the algorithm itself can properly discern motion. As it stands, the algorithm simply outputs the centroid data, and motion is determined by human review. In a clinical context, it would be more efficient for the algorithm itself to recognize and report motions. Therefore, a statistical threshold (e.g. “consecutive centroids that fall outside of 3 standard deviations”) needs to be designed for the algorithm to report motion.

To address the inability of the centroid technique to decisively characterize rotational motion, a variation on the centroid technique presented in this work is being developed. This technique involves dividing the PET scanner volume into segments and tallying the LOR midpoints found in each segment. Changes in these tallies over time could theoretically indicate motion. A crude example of angular segments in a transaxial plane is shown in Figure 53.
Figure 53: Segmentation technique. The scanner volume is segmented, and numbers of LOR midpoint in each segment are counted. In this simple example, it can be shown that a 90° roll motion occurred.

Another future development is the extension of the centroid calculation from a post-imaging measure to a real-time process. This would require feeding the live coincidence events, in place of the list-mode file, into the algorithm. Beyond increasing the flexibility of this technique, real-time centroid calculations would serve as a valuable tool for technologists. For example, they could potentially allow for mid-scan motion warnings. As such, real-time centroid calculations is one frontier for this work.

A final future development is to add a motion correction step to the algorithm. Once the motion characterization has been fine-tuned, the algorithm can either rebin the LORs, as proposed by Menke et al. and Daube-Witherspoon et al.,$^{6,10}$ or determine the appropriate transformation matrices to realign post-motion acquisition frames, as described by Picard and Thompson.$^{8}$ Including motion correction in the algorithm is a final major step in making the algorithm robust enough for clinical application.
4.5 Applications

The centroid technique presented in this work has a wide variety of potential uses. At minimum, it can be used as a motion detector to warn PET technologists and radiologists that there was motion during a scan, without any further analysis. This alone can prove valuable, particularly as a real-time alert. A more involved use would be to reveal when during the scan the motions took place. This could help determine which portions of a scan to keep, or how to divide the acquisition into motion-free image frames. In both of these uses, the algorithm would serve to solely detect motion, and nothing further.

A next potential use would be to use the algorithm to characterize motion. Along with the times of motion, the algorithm would determine the types of motion that occurred and the amounts. With that information, manual image registration could be performed on separate image frames.

A final use of this technique would be to use it for automated motion correction. The algorithm would be run on a list-mode file (or in real-time), and the result would be a motion-free image.
The range of potential applications for the algorithm is demonstrated in the flow chart in Figure 54.

**Figure 54: Algorithm potential applications flow chart.**

Earlier, a clinical example of a PET brain image with motion artifacts was shown (Figure 5). The list-mode file for this scan was run through the algorithm. Five time points with discrete motions were identified. The scan was divided into six separate acquisition frames with these five time points, and each frame was reconstructed as a separate image. The first of these images was displayed alongside the original image in Figure 5. All six images are shown in Figure 55.
Figure 55: PET brain scan from Figure 5. Motion was detected at five distinct time points. The top image was reconstructed with the entire raw data file. The following six images were reconstructed with data acquired between separate motions. Images courtesy of Timothy Turkington, Ph.D.
5. Conclusion

A data-driven technique for motion detection and characterization in PET brain scans has been developed. A program was written to calculate radioactivity centroids from LOR midpoints in list-mode data. Changes in these radioactivity centroids correlate linearly with mid-scan changes in head orientation and position. 1-s centroid measurements taken on scans with ~1mCi of FDG with no motion present have standard deviations of 0.026 cm in the x- and y-dimensions and 0.020 cm in the z-dimension, indicating that these centroids can represent discrete motions on the submillimeter scale. While the radioactivity centroid can precisely indicate the amount of motion during an acquisition, it fails to distinguish what type of motion occurred.
References


