Asymmetric Cone-Beam Transmission Tomography

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Abstract

Transmission scans are an important part of the process of obtaining patient attenuation maps for accurate SPECT (single photon emission computed tomography) imaging. In order to avoid truncation of the projection data, asymmetric cone-beam imaging geometries were investigated. By using the Beacon\textsuperscript{TM} SPECT system, it was shown that in the acquisition process transmission and emission imaging geometries can be different. Utilizing this new system, cone-beam transmission scans can be performed while the parallel collimators are attached for emission data acquisition. Patient and phantom studies were performed and compared with x-ray CT. A whole-body helical scan to obtain a whole-body attenuation map is also proposed. For helical asymmetric cone-beam geometries, at least two detectors are required in order to provide sufficient cone-beam data. If two detectors are used, they must have opposite geometries. Computer simulations were performed to demonstrate the feasibility of the proposed asymmetric helical cone-beam scanning geometry.

I. INTRODUCTION

In SPECT, an attenuation map is required for accurate emission attenuation correction. Usually the attenuation map is formed from transmission data. There are many transmission data acquisition methods that can be used including a flood source [1], a scanning line source with parallel-beam geometry [2], a scanning point source with fan-beam and offset fan-beam geometry [3], or a fixed line source with fan-beam geometry [4]. In PET, a moving point source has also been used to obtain transmission information [5]. The use of a scanning source is more complicated than the use of a fixed source because of the additional mechanics and electronics required for scanning, which can increase cost and potentially produce an additional factor that may fail during imaging.

Scanning sources are at an advantage when scanning is combined with an electronic window. With this combination down scatter fraction from emission to transmission can be reduced by a factor larger than ten. A point source is preferred over a line source or a sheet source because the patient will receive less radiation exposure. Additionally, the amount of activity for a scanning point source can be on an order of magnitude lower than a scanning line source.

If a fixed transmission point source is used, the necessity of using a cone-beam imaging geometry is a direct consequence. However, when using a cone-beam geometry there are two significant problems that must be addressed. The first problem is that the projection data of an object will most likely be truncated due to the converging beam geometry. The second problem is that a planar scanning orbit does not provide a complete data set for a stable, artifact-free reconstruction.

The use of offset cone-beam, i.e., asymmetric cone-beam, collimators has been suggested for combined emission/transmission brain SPECT [6]. By using an asymmetric cone-beam geometry, as illustrated in Fig. 1, the truncation of projection data can be avoided if the axis of rotation and one side of the patient boundary are in view of the detector. In this paper, we employ the asymmetric cone-beam geometry to overcome the truncation problem of cone-beam transmission projections.

If the planar scanning orbit is replaced by a non-planar orbit, for example a helical orbit, under certain conditions [10] a complete cone-beam projection data set can be obtained. In

![Figure 1. Illustration of asymmetric cone-beam geometry. The detector is set wide enough to ensure that at least half of the object from one edge to the center of rotation is in the field of view.](image-url)
In this paper, a computer simulation is used to illustrate the application of a helical orbit in asymmetric cone-beam tomography.

In this study the Beacon™ SPECT system is used. Recently, Picker Inc. introduced the Beacon™ SPECT system (see Fig. 2) which can acquire transmission and emission data simultaneously. The transmission source has enough energy to penetrate the collimator, which makes it possible to use different imaging geometries for transmission and emission data acquisition. In clinical application Beacon™ is designed to perform fan-beam transmission scans with scanning point sources and scanning electronic data acquisition windows. In this paper, we use a cone-beam transmission data acquisition with fixed point sources. In practice this does not prevent us from using a cone-beam geometry for transmission imaging while using a parallel-beam geometry for emission imaging at the same time.

II. METHODS

A. Beacon™ Cone-Beam Studies

A picture of the Beacon™ system (Picker's IRIX) is shown in Fig. 2. The configuration shown includes a Barium-133 point source of 10 mCi (370 MBq) mounted on the edge of detectors #1 and #2. Detector #3 was not used for transmission data acquisition. Ba-133 has a half-life of approximately 10.5 years and has a main energy peak at 356 keV and two small energy peaks at 383 keV and 302 keV, respectively. Low-energy (140 keV) high-resolution (LEHR) parallel collimators were mounted for the emission scan. The LEHR collimators had a hole size of 1.22 mm, a hole length of 27 mm, and a septum thickness of 2.03 mm. The high energy transmission photons penetrated the collimator septa, forming an asymmetric cone-beam imaging geometry (see Fig. 3). The asymmetric cone of transmission rays from each source covered slightly more than half of the patient's torso.

This Beacon™ system can also be used in an alternative transmission/emission SPECT study, in which asymmetric cone-beam transmission data and parallel-beam emission data are acquired non-simultaneously. At each projection stop, transmission and emission data are acquired sequentially. When the emission data are acquired, the transmission source shutter is closed. Assuming that the emission photon energy range is between 70 and 140 keV, which is much lower than the transmission photon energy (356 keV), the transmission data are not affected by the emission photons.

We have applied the Beacon™ cone-beam transmission data acquisition method to other parallel collimators, including medium-energy general all-purpose (MEGAP) and 511 keV axial collimators. The 511 keV collimator is made of parallel lead slits. Reconstructions using projection data from only one detector has also been performed.

In these reconstructions, a planar scanning orbit is used. It is well known that this orbit does not provide a complete data set that enables artifact-free reconstruction, especially in the regions far away from the orbit plane. Fortunately, we have a relatively long focal length, and the artifacts due to cone-beam data-incomplete sampling are small and do not detract from the quality of the reconstruction. Another fact one needs to keep in mind is that the cone-beam data-incomplete artifacts are much less pronounced when using an iterative reconstruction than when using a filtered backprojection reconstruction [9].
A.1. A patient Beacon™ study

In a patient study, the detectors rotated 360° with 120 stops. The data acquisition time was 1 second per view. Data were acquired in 128 × 128 arrays, of which only 40 rows were illuminated by the point source due to the small point-source collimator angle. The detector pixel size was 4.669 mm. The cone-beam focal length was 82 cm, and the focal point was shifted by 14.94 cm towards the edge of the collimator. The angle between the two detectors that acquired projection data was 78°. The detectors rotated in a noncircular, planar orbit around the patient. After the patient was removed from the scanner, a one minute reference scan was acquired, and was used to convert the transmission data into line-integrals.

After the patient transmission projections were converted into line-integrals, a cone-beam iterative (emission) ML-EM (maximum likelihood expectation maximization) algorithm [8] was used to reconstruct the image (100 iterations). Our previous study shows that the emission ML-EM algorithm results in better images than the transmission ML-EM algorithm in attenuation map reconstruction. The instability of the transmission ML-EM algorithm is partially caused by the approximation used during the derivation of the algorithm.

The patient scanned was a 134 pound male. Larger patients would require a longer scanning time.

A.2. A phantom Beacon™ study with LEHR collimators

A torso phantom was used in a cone-beam Beacon™ study with LEHR (low energy high resolution) parallel collimators. The data acquisition parameters were the same as in the patient study, except that the transmission data were acquired for 3 seconds per view, and the reference scan was 2 minutes long. The images were reconstructed with 100 iterations of emission using the ML-EM algorithm.

A.3. A phantom Beacon™ study with 511 keV axial collimators

The same torso phantom used in A.2 was used in another cone-beam Beacon™ study with 511 keV parallel axial collimators, which were designed for positron coincident detection (PCD) data acquisition. The data acquisition setup was the same as in A.2 except that the transmission data were acquired in 60 stops over 360°, with 8 seconds per view. A one-minute reference scan was also acquired. Image
reconstruction was performed using 100 iterations of the ML-EM algorithm.

A.4. X-ray CT study

A Picker PQ-6000 CT scanner was used to scan a torso phantom. The x-ray CT study was used as the gold standard against which other nuclear medicine studies were to be compared. The transaxial resolution was 1 mm and the slice thickness was 1 cm in the x-ray CT study. The scan was performed with 200 mA and 100 kV. The acquisition time was 1 second per slice. The CT images were reconstructed in 512 x 512 arrays, which were subsequently collapsed to 128 x 128.

B. Asymmetric Cone-Beam Helical Scans

In this section, we outline the advantages of using a helical scanning orbit (see Fig. 5) to obtain a whole-body attenuation map. An orbit is referred to as the point source trajectory in a scan. In a helical scan the orbit pitch requirement is crucial [10]. In a previous study, the pitch requirement was determined by Palamodov’s data sufficiency condition [9].

Palamodov’s condition can be stated as every plane that passes through an object must contain a line that is measured in at least one projection. Using this condition, it can be seen that a single asymmetric, cone-beam geometry does not have the capacity to provide sufficient data with a helical scanning orbit. Thus for our application, two asymmetric cone-beam detectors were used.

In this application, if the focal-point of one detector is off-centered to the left, then the focal-point of the other detector must be off-centered to the right (see Fig. 6). Figure 7 illustrates that untruncated projection data can be formed by combining data from detectors #1 and #2 when they have the same point source position. Therefore, the design of the whole-body helical asymmetric cone-beam system requires at least two detectors. The third detector, when used, can be dedicated to emission data acquisition.

The detectors can be arranged in an L-shape (90° apart) as depicted in Fig. 7. The angle between the two detectors can
Figure 12. Patient cone-beam transmission data reconstruction. Projections were acquired with a Beacon\textsuperscript{TM} system with LEHR parallel collimators attached.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{slices.png}
\caption{Slices #53 to #76}
\end{figure}

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{slices.png}
\caption{Slices #56, #64, and #72}
\end{figure}

\section{III. RESULTS}

\subsection{A. Beacon\textsuperscript{TM} Cone-Beam Studies}

The transmission projections were acquired using asymmetric cone-beam geometry. Due to the point-source collimation, the cone-angle in the axial direction was smaller than the cone-angle in the transaxial direction. In a typical scan, approximately 40 rows of pixels (about 18 cm) on the detector were illuminated. Figures 10 and 11 show a projection view in a torso phantom study with a LEHR collimator and a PCD collimator, respectively. The brightest spot in the projection image was caused by the photons that passed directly through the collimator holes, indicating the location of the point source. This bright spot also appears in the blank scans, which were taken after the subject (patient or phantom) was removed. The normalized projections, which are the ratio...
Figure 13. Torso phantom transmission x-ray CT study and cone-beam Beacon™ studies in which LEHR and PCD collimators are used. Profiles are drawn along the horizontal line shown in the corresponding images.
Figure 14. Three central cuts of an ML-EM reconstruction of a long Defrise phantom after 100 iterations. Detector size = 64x64. Image array size = 64x64x320. In the profiles, the solid lines are for the reconstructed image and the dotted lines are the ideal.

If the blank scan and the patient/phantom scan, do not have the bright spot.

Figure 12 shows the patient cone-beam transmission data reconstruction as described in Section II.A.1. Slices #53 to #76 are displayed in 128 x 128 images, and the cardiac region is within the range of these slices. Larger views of the top, mid, and bottom images are also shown in Figure 12 with the same pixel size. If more slices in an attenuation map are required, one can use a different source-collimator that has a larger cone-angle in the axial direction.

Figure 13 shows the reconstructions of the attenuation map for the torso phantom studies. An x-ray CT study (i.e., the gold standard) and three asymmetric cone-beam studies are presented. The first asymmetric cone-beam study was performed with the two-detector, LEHR-collimator, at 3 seconds per view, and 120 views per detector. The second study was the same as the first, except that only projection data from detector #1 were used in the reconstruction and projection data from detector #2 were not used. The third study was performed using the two-detector, PCD-collimator, at 8 seconds per view, and 60 views per detector. The study with PCD-collimators had more photon counts than the studies with LEHR-collimators. Thus, the images obtained from the PCD study are not as noisy as they are in the LEHR studies. Three slices (top, mid, and bottom) from each study were selected for display. A horizontal profile was drawn across the image array (i.e., 128 pixels) on each selected slice. In order to show the same relative phantom size, a sub-region of the 128 x 128 image array is displayed. Due to the poor resolution of SPECT cameras and the partial volume effect, the detail structure within the lungs cannot be seen in nuclear medicine studies, however they can be clearly observed in the x-ray CT study.

B. Asymmetric Cone-Beam Helical Scans

Figure 14 shows an ML-EM reconstruction of computer simulated data. Two sagittal cuts and two transaxial cuts are displayed. Severe artifacts can be seen at the top and bottom of the phantom, where the projection data are unavailable. However, the central region, where the projection data are
complete, is accurately reconstructed as indicated by the profiles.

IV. DISCUSSION

When using Picker’s Beacon™ SPECT system, the emission data acquisition geometry can be different from the transmission data. With parallel collimators attached, a fixed Ba-133 point source can be used to acquire asymmetric cone-beam transmission data. Fixed point sources reduce complications in transmission imaging. The truncation problem is avoided by using an asymmetric cone-beam geometry requires no moving parts are required.

Both parallel low-energy high-resolution and parallel 511 keV axial collimators have been used in patient and phantom transmission studies. The low-energy high-resolution collimator absorbs approximately 76% of gamma photons. Therefore, approximately 24% of the Ba-133 gamma rays that travel through the patient (in the 35% 356 keV energy window) reach the detector crystal. The Lead (Pb) x-rays generated from the collimator are lower than those in the Ba-133 356 keV window. There is also Compton scatter within the collimator. The effect of the Compton scatter in the collimator still needs to be carefully investigated.

We also attempted to reconstruct images with projection data collected by only one detector. As expected, the reconstruction using data from one detector is noisier than the reconstruction using data from two detectors. The data set acquired by one detector has a more severe problem due to the incomplete sampling of the cone-beam data. There is a potential for image distortion which can occur in regions far from the orbit-plane.

A helical SPECT scan is suggested to acquire comprehensive data for whole-body attenuation map reconstruction. A helical orbit is able to provide a complete cone-beam data set and the image distortion, due to data-incompleteness, can be eliminated. For asymmetric cone-beam geometries, at least two detectors must be used, and they must form two opposite asymmetric cone-beam geometries. Our proposed whole-body helical scan will have many applications such as bone and spinal imaging, as well as prostate cancer imaging.

Even though iterative reconstruction algorithms are used to reconstruct the images, analytical reconstruction algorithms can also be used. In the planar orbit situation, an iterative algorithm is preferred over an analytical method because it produces much fewer image distortion artifacts. In the helical orbit situation, an analytical method for asymmetric cone-beam geometry is under development.

ACKNOWLEDGMENTS

We thank Roger Roberts for acquiring x-ray CT data in this study. Also, we thank Sean Webb for editing the manuscript.

REFERENCES


