Published in final edited form as:

IEEE Trans Nucl Sci. 2008 October 1; 55(5): 2518–2526. doi:10.1109/TNS.2008.2003255.

Quantitative Evaluation of Half-Cone-Beam Scan Paths in Triple-Camera Brain SPECT

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Abstract

In this study related to human brain SPECT imaging, simulation of half-cone-beam (HCB) collimation with different scan paths is performed and compared with simulated fan-beam and parallel-hole circular orbit acquisitions of disk-phantom projection data. Acquisition types are quantitatively evaluated based on the photon detection efficiency, the root-mean-squared error, contrast and signal-to-noise ratio measurements of the reconstructed images. We demonstrate that a triple-camera SPECT system with half-cone-beam collimators and circle-and-helix scan paths can offer up to a 26% efficiency increase over fan-beam, and up to a 128% increase over parallel-hole collimators for equal spatial resolutions, and display no visible axial sampling artifacts in reconstructed disk-phantom images. In addition, we perform qualitative experimental evaluation of triple-HCB circle-and-helix acquisition using a Hoffman 3D brain phantom. Reconstructed brain phantom images show improved quality due to reduced noise and no apparent sampling artifacts. Triple-HCB circle-and-helix SPECT has a potential for improved brain imaging, producing higher image quality with a smaller reconstruction error and better lesion detectability due to increased efficiency for equal spatial resolution compared to conventional fan-beam and parallel-hole SPECT.

Keywords

axial sampling; brain SPECT; circle-and-helix; half-cone-beam; helical-path; triple-camera

I. INTRODUCTION

High sensitivity half-cone-beam (HCB) collimators were designed by Jaszczak et al. [1], [2] specifically for brain SPECT imaging. As compared to full-cone-beam (FCB) collimation, HCB collimation allows close positioning of the γ -camera to the patient's head providing

improved spatial resolution and avoiding truncation of the caudal part of the brain due to camera interference with the patient's shoulders, as demonstrated in Fig. 1. For a given spatial resolution, the HCB-collimated camera efficiency is superior to that of parallel-hole beam (PB) or fan-beam (FB) collimated cameras. However, a conventional circular-orbit (CO) acquisition using HCB (or FCB) collimation produces axial artifacts due to insufficient axial sampling. Only the slice located on the transaxial focal plane of the HCB collimator, defined as a focal plane perpendicular to the camera's axis of rotation (AOR) and the collimator surface, is sufficiently sampled during a CO acquisition. To alleviate these axial distortions, helical HCB trajectories have been investigated [3], [4] similar to helical SPECT proposed for long-object imaging [5], [6], pinhole [7] and X-ray CT scans [8]. Helical SPECT scans are typically implemented by translating the patient axially as the gantry revolves in a circular orbit. The combined motion of the patient translation and the gantry rotation results in a helical trajectory of the collimator focal point around the patient.

Although helical scan paths markedly reduce sampling artifacts in HCB brain SPECT [3], [4], it is unclear whether they are beneficial overall, since the patient translation quickly moves the caudal portion of the brain outside the camera's field of view (FOV), thereby reducing camera/trajectory efficiency for the caudal portion. Essentially, there is a tradeoff between the better efficiency of HCB CO trajectories and the better sampling of helical HCB trajectories. In the present paper, we assess camera/trajectory efficiencies quantitatively and investigate scan paths which involve a combination of CO and helical components. The hypothesis of the present paper is that circle-and-helix scan paths of HCB-collimated triple-camera (triple-HCB) SPECT systems can provide better photon detection efficiency and better signal-to-noise ratio (SNR) than do PB and FB-collimated triple-camera systems (at a given spatial resolution), while at the same time avoiding any apparent incomplete-sampling artifacts.

II. METHODS: SIMULATION AND EXPERIMENT

A. Simulation

A1. Phantom and Collimator Parameters—A disk phantom was used in simulations for revealing evidence of insufficient axial sampling during HCB CO acquisition. The phantom consisted of 5 hot and 4 cold disks (thickness = 1.44 cm, radius = 9.0 cm) perpendicular to the AOR. The total length of the phantom, centered on the AOR, was 12.96 cm. The dimensions of the phantom were chosen to generally resemble a human head. Each hot disk contained 1 hot rod ($r_0=0.72$ cm) in the center, and 4 cold rods ($r_1=0.36$ cm, $r_2=0.72$ cm, $r_3=1.08$ cm, $r_4=1.44$ cm) parallel to the AOR, located 6.84 cm away from the center of the disk and separated by 90° (see Fig. 2). The cold disks and rods had zero activity. The hot rod had 50% more activity than the hot disks. The FOV of the γ -camera was chosen to be similar to the general-purpose triple-camera SPECT system (Triad XLT, Trionix Research Laboratories, Twinsburg, OH) at the Duke University Medical Center (DUMC) SPECT Research Laboratory, measuring 46.08 cm transaxially and 23.04 cm axially. The three cameras were separated by 120° .

For the purpose of efficiency calculation using MC simulations collimator geometric parameters such as thickness, the hole size and shape, and septal thickness were modeled. The HCB and FB collimators were 4.0 cm thick with 0.145 cm square hole size and 0.022 cm septal thickness. Parameters of the simulated PB collimator were adjusted so that the resolutions of HCB, FB and PB collimators in object space were approximately equal. Namely, 0.132 cm square hole size was used for the PB collimator, considering the detector intrinsic resolution of 0.35 cm. Spatial resolutions were calculated for a point source located 13.0 cm from the collimators, which corresponds to the radius of rotation (ROR) of the γ -camera. The focal lengths of the HCB and FB collimators were 50.0 cm measured from the patient side of the collimator. The focal point of the HCB collimator was shifted axially by 11.52 cm towards the base of the phantom as measured from the center of the camera.

A2. Data Projection and Reconstruction—Projection data were simulated on a 256×128 grid using a ray-driven forward-projection algorithm. Iterative reconstruction of the phantom was performed using an ordered-subsets expectation-maximization (OSEM) algorithm [9] with 18 subsets, 10 iterations for noise-free and 5 iterations for noisy projection data. Image voxels were cubical with an edge length of 0.18 cm. Collimator resolution was not modeled in ray-driven forward-projection and OSEM reconstruction algorithms. Detector intrinsic resolution of 0.35 cm full width at half maximum (FWHM), modeled as Gaussian, was used only in noisy projection data. Attenuation was modeled using a uniform map based on phantom dimensions. The attenuation coefficient was $\mu = 0.15 \text{ cm}^{-1}$. Scatter and septal penetration effects were not modeled in forward-projection and reconstruction algorithms.

Noisy projection data were obtained by sampling from Poisson distributions with mean values given by simulated noise-free projection data. The mean value of Poisson distribution was 2.7×10^6 counts per camera utilizing a FB collimator, which is what we would anticipate during a typical patient brain scan using a triple-camera SPECT system at DUMC. Projection data for HCB and PB acquisitions were scaled according to their (relative to FB) efficiencies. Reconstructed noisy images were post-filtered using Hann filter with 1.4 cycles/cm cutoff frequency.

A3. Data Acquisition Trajectories—Photon detection efficiency and object sampling may be the two most important parameters to optimize in data acquisition, provided spatial resolution is adjusted to be similar in each scan. Sampling is improved by covering all spatial frequencies of the object. Efficiency, defined as a ratio between the number of photons detected and emitted by the source, may be increased by using a convergent-beam collimator and having the object in the FOV throughout the scan.

Both PB and FB collimators offer sufficient axial sampling with a CO acquisition, and FB offers higher efficiency. HCB collimators are even more efficient than FB due to their axially-convergent geometry, but a CO acquisition causes insufficient axial sampling artifacts. The sampling is markedly improved using helical scan paths, but the projections of the brain become truncated shortly after the start of a helical scan. As a result, the efficiency drops. To avoid low efficiency, especially in the caudal part of the brain, we considered using “circle-and-helix” trajectories with HCB collimation. Here, the system rotates 360° in a circular orbit with a motionless phantom located close to the edge of the FOV. The phantom is then translated across the FOV in the caudal direction as the cameras continue rotating (see Fig. 3). We study the trade-off between axial sampling and the efficiency by varying the amount of views per circular and helical parts of the scan, assuming equal time per view and equal total scan time. Increasing the number of axial views improves axial sampling, but decreases the efficiency. The acquisition trajectories for a triple-camera SPECT investigated in this article are listed in Table I. Alternative variations of HCB circle-and-helix scan paths with variable number of views and time per view are discussed in Section IV.

A4. Quantitative Evaluation—Collimator performance for each scan path was evaluated based on photon detection efficiency computation, deviation of the reconstructed hot and cold disks from their true (bitmap) images, contrast and SNR measurements of the hot and cold rods in the disk phantom.

A4.1. Photon detection efficiency: Photon detection efficiencies were calculated using Monte Carlo simulation of the photon transport, which has several advantages over analytical calculation of efficiency [10], including accurate estimations of Compton scatter and attenuation effects in the object, and correct accounting of lost counts due to the object truncation. Scatter and attenuation were simulated using attenuation coefficient for water $\mu = 0.15 \text{ cm}^{-1}$. Scattering angle was simulated using the Klein-Nishina formula for differential

cross section with respect to solid angle of scattering. The energy of the scattered photon was computed from the scattering angle. Scattered photons with energies less than 126 keV, which corresponds to the threshold of the $\pm 10\%$ energy window of Tc-99m photopeak centered at 140 keV, were rejected. The uncertainty of efficiency calculation was (1.0-1.5) %, based on the counts of MC simulated and detected photons. The effects of septal penetration and photopeak detection efficiency were not simulated, since they are mostly independent of the collimator type and a scan path.

A4.2. Noise-free projection data: Reconstructed noise-free image quality can be evaluated quantitatively in comparison with the true bitmap image, if it is known *a priori*, in terms of pixel-by-pixel root-mean-squared error (RMSE),

$$\delta = \sqrt{\frac{1}{n} \sum_{i=1}^n (f_{\text{true}}^i - f_{\text{reco}}^i)^2}, \quad (1)$$

where f_{true}^i and f_{reco}^i are activities in i -th pixel of the true and reconstructed image slices, and n is the total number of pixels in a 2D slice. The values of the reconstruction error δ for each slice are then averaged over the disk containing eight slices. Large values of δ indicate distorted reconstructed image of the phantom. The detector intrinsic resolution was not taken into account during the δ calculation to avoid possible non-sampling distortions affecting the RMSE.

Reconstruction error in terms of the RMSE is a global evaluation of the reconstructed disks and the phantom. To effectively address localized image evaluation, like lesion visibility and detectability, contrasts of the hot and cold rods were measured. Contrast is defined as

$$C = (f_{\text{lesion}} - f_{\text{bkg}}) / f_{\text{bkg}}, \quad (2)$$

where f_{lesion} and f_{bkg} are the activities per pixel in the lesion (here, the rods mimic lesions) and the background (normal tissue) in a 2D slice. Background activity was calculated from a circle around the lesion, excluding the lesion activity.

A4.3. Noisy projection data: Poisson noise is inevitable in acquired projection data. The level of noise directly depends on the number of photons detected, which is related to the photon detection efficiency discussed above. Reconstructed and post-filtered image quality and lesion detectability were evaluated using SNR for hot and cold rods, averaged from an ensemble of $N=16$ noisy realizations:

$$\text{SNR} = \frac{\frac{1}{N} \sum_{i=1}^N (|g_i^{\text{signal}} - g_i^{\text{bkg}}|)}{\frac{1}{N} \sum_{i=1}^N \sigma_i}, \quad (3)$$

$$\sigma_i = \sqrt{\frac{1}{M} \sum_{j=1}^M (g_j^{\text{bkg}} - g_j^{\text{bkg, noise-free}})^2}$$

Here, g_i^{signal} and g_i^{bkg} are the normalized activities in the rod and a larger circle around it ($r_{\text{bkg}} = 2.3 r_{\text{signal}}$) from i -th noisy realization, σ_i is the background noise calculated from pixel-by-pixel background variance for $j=1, \dots, M$ background pixels. “Noise-free” images are reconstructed from very high count ($\sim 3 \times 10^9$ counts per camera) projection data and are

subtracted from the noisy images to determine the noise. Larger SNR values correspond to better lesion detectability.

B. Experiment

B1. Data acquisition system—A triple-camera Triad-XLT SPECT system was used to acquire HCB circle-and-helix, FB and PB circular-orbit projection data. Collimator specifications are shown in Table II. In this table, the spatial resolutions (full-width-at-half-maximum (FWHM)) were measured in object space. Both sensitivity and resolution were measured near the transaxial focal plane at a distance of 12.5 cm from the patient side of the collimator. Decreased sensitivity of the available FB collimator due to slightly better resolution compared to the HCB collimator is partially compensated with sensitivity gain due to shorter FB focal length. The HCB collimator had its focal point shifted axially 12.1 cm towards the base of the brain. The transaxial focal plane of the HCB collimator was unintentionally located about 6 mm beyond the axial extent of the scintillation camera's field of view.

B2. Experimental phantom—A Hoffman 3D brain phantom (Model BR/3D/P, Data Spectrum Corp., Hillsborough, NC 27278-2300) was used in the experimental study. The phantom models 19 different transaxial brain slices. Each brain slice is ~6.4 mm thick and consists of five polycarbonate sub-layers. The maximum trans-lateral dimension of the brain is ~13 cm. The maximum anterior-posterior dimension is ~18 cm. The maximum cranial-caudal dimension is ~12 cm. The 3D brain phantom was filled with a solution of water containing $6.9 \mu\text{Ci}\cdot\text{cm}^{-3}$ Tc-99m and placed on the phantom holder of the patient bed. The (grey matter):(white matter):ventricle uptake ratio of 4:1:0 is obtained using a partial volume effect when imaging the five sub-layers that make up a single brain slice. The 19 brain slices were placed in a polymethyl methacrylate (PMMA) cylinder measuring ~17.5 cm in height and ~20.5 cm diameter. There was no activity in the cylinder outside of the brain.

B3. Data acquisition and image reconstruction—Triple-camera circular-orbit data were acquired using continuous gantry rotation with 1° sampling over 360° . To resemble clinical patient brain scans at DUMC, scan time was adjusted to acquire ~3 million counts per camera utilizing the FB collimator. Circle-and-helix scans with HCB collimation were performed in two steps: a 240° circular orbit with 1° sampling, followed by additional 240° gantry revolution with 2° sampling plus continuous phantom translation in the axial direction for a combined helical scan. The acquisition time for the circular-orbit portion of the scan was twice as long as the helical portion, mimicking acquisition type (c) from Table I. During the helical scan the phantom traveled 14.2 cm and exited the camera's FOV. The camera's ROR was 13 cm. We currently have only one HCB collimator available in our laboratory, therefore multiple single-detector acquisitions were used to produce the projections that would have been obtained with a triple-HCB collimator configuration. The three detectors were equally spaced by 120° . At any instant during a true triple-HCB circle-and-helix acquisition, the respective transaxial focal planes of all three cameras would be coplanar. Throughout the experiment, the scan times were adjusted for Tc-99m decay. The adjusted scan times were equal for the HCB, FB and PB scans.

A 20% (FWHM) energy window centered at 140 keV was used for all acquisitions. The acquisition matrix size was 256×128 (0.18 cm pixel size). Images were reconstructed iteratively using OSEM algorithm with 24 subsets and 2 iterations. Attenuation was modeled using a uniform map based on phantom dimensions. The attenuation coefficient was $\mu = 0.12 \text{ cm}^{-1}$. Scatter and collimator point-spread response were not modeled. Reconstructed images were post-filtered using a Hann filter with 2.5 cycles/cm cutoff frequency.

III. RESULTS

A. Simulation

A1. Photon detection efficiency—Due to axially converging geometry of HCB collimators and specific scan paths discussed in this article, photons from different disks in the phantom are detected with different efficiencies. Efficiencies for each hot disk of the phantom with and without scatter and attenuation effects were calculated using MC simulation for specific collimator types and scan paths (see Fig. 4). During a helical scan caudal disks provide less photon counts than do cranial disks, which are in the FOV throughout the scan. The contrary is true for HCB CO acquisition, where the caudal disk photons are detected with the highest efficiency. Photon attenuation also contributes, decreasing the overall efficiency by $\sim 50\%$, but hardly affecting the most cranial disk. Lack of attenuation above the cranial disk of the simulated phantom is the primary reason for the efficiency “jump” between disks 4 and 5 observed in Fig. 4 (bottom plot) for HCB scans. HCB CO acquisition offers the highest photon detection efficiency. The second most efficient acquisition type is HCB circle-and-helix, with 240 circular and 120 helical views (assuming equal time per view). The advantage of every other HCB acquisition over the FB is only for the cranial part of the phantom. Table III presents photon detection efficiencies per camera averaged over five hot disks.

A2. Reconstructed noise-free image quality—RMSE values of the reconstructed hot and cold disks are presented in Table III in terms of percentages of the true mean values of the hot disk activity. Axial distortions with HCB CO acquisition are quantitatively evident. The RMSE from other scan paths are substantially smaller and can be tolerated for all practical purposes. Non-zero activity in cold disks for HCB acquisitions is evidence of a slice-to-slice cross talk artifact [11] generally present in cone-beam acquisitions with insufficient axial sampling. With helical scan paths this artifact is significantly reduced and does not affect the image quality noticeably. Noise-free images of the disks had no visible artifacts and were practically indistinguishable with HCB circle-and-helix, FB and PB CO acquisitions.

Average contrast values with respect to true contrasts, $C_{\text{reco}}/C_{\text{true}}$, with HCB circle-and-helix acquisitions were 96% for hot rod r_0 , 86% and 92% for cold rods r_1 and r_2 . In comparison, the three rod contrasts were 94%, 82% and 90% with FB, and 93%, 80% and 89% with PB CO acquisitions. The HCB contrasts are slightly better than FB and PB due to higher convergence rate of the iterative algorithm.

Following both the efficiency and the RMSE calculations of different HCB scan paths we conclude that HCB acquisition type (c) from section II.A offers an excellent trade-off between axial sampling and efficiency. In the following sections we will only consider this particular HCB acquisition type and compare it with FB and PB CO acquisitions.

A3. Reconstructions with Poisson noise—Figs. 5(a-c) show reconstructed and post-filtered images of the centrally located transaxial slices of the five hot disks with the HCB, FB and PB collimations. Images from the HCB acquisition (Fig. 5a) shown in a caudal-to-cranial order clearly demonstrate improved quality of the cranial disks due to the increasing photon detection efficiency. All rods are visible in transaxial, coronal (Fig. 5d) and sagittal (Fig. 5e) slices. Both coronal and sagittal slices demonstrate no visible axial artifacts.

Reconstructed FB images (Fig. 5b) are, on average, noisier compared to the images from the HCB acquisition due to the lower efficiency. As expected, PB reconstructions are the noisiest of the three acquisition types (Fig. 5c). Rods r_0 and r_1 are hardly visible in some disks. The advantage of the HCB acquisition over FB and PB was confirmed quantitatively by the SNR study of rods r_0 , r_1 and r_2 (see Table V). The HCB SNR values were the largest of the three collimator types, except for the caudal most disk for which, as expected, they were similar to

the FB SNR. Moreover, the HCB SNR increased for cranial disks due to increasing efficiency, and was always greater than 3, which means the lesions (rods) were always detectable. The PB SNR values for rods r_0 and r_I were below 3, showing poor lesion detectability.

The uncertainties in the average SNRs ($\sigma_{\langle SNR \rangle}$) in Table V were calculated as follows:

$$\sigma_{\langle SNR \rangle} = \frac{\sqrt{\frac{1}{N-1} \sum_{i=1}^N (SNR_i - \langle SNR \rangle)^2}}{\sqrt{N}} \quad (4)$$

for $N=16$ noisy realizations, assuming Gaussian distribution of the SNR values. The errors can be reduced by using a larger number of noisy realizations.

B. Experiment

The number of photons detected during the HCB circle-and-helix, FB and PB circular-orbit scans described in section II.B are shown in Table VI. Efficiency gains of 37% and 93% were achieved with HCB circle-and-helix scan compared to FB and PB CO acquisitions correspondingly. The total number of photons acquired during the FB CO scan was similar to the count rate anticipated during a typical patient brain scan. Reconstructed and post-filtered brain phantom images are shown in Fig. 6, along with the bitmap images shown as a reference. Qualitative evaluation of the images and their profiles showed reduced noise in HCB reconstructed images compared to images from FB and PB-acquired data. No apparent HCB axial sampling artifacts were observed.

IV. DISCUSSION

In this paper, we evaluated different acquisition types for improved human brain imaging involving HCB, FB and PB collimators in triple-camera SPECT. HCB collimators have an advantage of higher sensitivity and close positioning to the patient's head without truncation of the caudal part of the brain due to interference with the patient's shoulders. Circular orbit acquisition with HCB collimators is known to cause axial distortions due to insufficient axial sampling of the brain. Helical scan paths markedly improve axial sampling, but cause decreased photon detection efficiency for the caudal part of the brain, since it exits the camera's field of view shortly after initiating the patient translation. To increase detection efficiency by maintaining the brain (especially its caudal part) within the FOV we propose using a circle-and-helix scan path, where the circular orbit scan is followed by the patient translation. By increasing the number of circular-orbit views and reducing the number of helical (axial) views we increase the efficiency but compromise axial sampling. Using simulations we demonstrate that HCB acquisition type (c) described in section II.A offers an excellent trade-off between the efficiency and axial sampling. For a 360-view triple-HCB scan it assumes 240 circular views and 120 helical (axial) views for equal view time. The number of views and time per view may be modified, but the ratio between CO scan time and helical scan time should remain constant. For instance, 240 circular views can be replaced with 120 views with twice the time per view, and 120 axial views can be replaced with 240 views with half the time per view. Equal total scan time requirement validates performance comparison between different acquisition types. Reducing the number of views may be beneficial during an experimental study with a "step-and-shoot" scanning protocol, where the detector does not register photons during the gantry rotation from one scanning view to another. For simulation purposes, or an experimental scan with continuous detection, the large number of views only affects reconstruction time.

Triple-HCB circle-and-helix acquisition described above proved to have the highest efficiency after HCB CO acquisition that was discarded due to strong axial artifacts observed in disk phantom reconstructions. The average HCB circle-and-helix efficiency over the five hot disks was 26% higher than FB CO, and 128% higher than PB CO acquisitions. To quantify reconstructed image quality, pixel-by-pixel RMSE with respect to their true bitmap images were computed for hot and cold disks of the phantom using noise-free projection data. HCB circle-and-helix errors were insignificant (less than 4% for hot, and ~1% for cold disks) and comparable to the FB and PB RMSE values. They did not affect the image quality considerably.

Reconstructions with Poisson noise and detector intrinsic resolution were performed to more closely resemble clinical brain studies. Quantitative analysis showed that HCB collimation with the circle-and-helix scan path outperforms both FB and PB CO acquisitions. HCB reconstructed disk images demonstrated reduced noise, improved quality and better lesion detectability. No visible axial artifacts were detected.

Experimental Hoffman 3D brain phantom study demonstrated performance and advantages of the HCB collimator in practice. Comparison between equal resolution HCB and PB acquisitions with equal scan time showed 93% efficiency gain with the HCB circle-and-helix scan. As a result, HCB reconstructed images had less noise and higher quality. A similar comparison in a disk phantom simulation study yielded a 128% HCB efficiency gain. The difference between the simulation and the experimental studies stems primarily from the phantom differences. A significant efficiency gain in a simulated disk phantom scan comes from the top hot disk, which is hardly affected by attenuation and has twice the efficiency of the bottom disk. In the brain phantom, however, the top slices are significantly smaller, and the “brain” is placed inside an attenuating cylinder, which substantially reduces contribution from the cranial part of the brain compared to the simulated disk phantom.

Triple-HCB data sufficiency

Mathematically exact 3D brain image reconstruction would require complete sampling of the brain. If sampling is insufficient, artifacts may be observed in the reconstructed images. Artifacts can vary from hardly visible to severe, depending on the extent of data insufficiency. In strictly mathematical terms experimental data are always “incomplete”, since they include physical factors such as limited spatial resolution, discrete sampling along the focal-point trajectory, and finite-size pixels. Actual physical SPECT imaging is further degraded by noise, attenuation, and scatter, and by the limited accuracy with which these effects can be modeled during image reconstruction. Accordingly, the primary emphasis of this paper was not to establish mathematically perfect complete sampling with HCB collimation, but was instead to demonstrate empirically that certain circle-and-helix scan paths can provide better quality (through noise reduction) SPECT images that have no apparent incomplete-sampling artifacts. That noted, several heuristic considerations have been crucial in designing such circle-and-helix trajectories [12]. Further investigation may be required to establish more rigorous criteria and methods leading to complete sampling with HCB tomography, including single and double-camera cases. Recent advances in image reconstruction techniques using 1D Hilbert transform inversion for truncated projection data [13]-[17] already offer promising results for parallel-hole and cone-beam tomography, and may be found to be applicable to HCB tomography as well.

V. CONCLUSION

In this paper we demonstrate that triple-camera half-cone-beam circle-and-helix acquisition in brain SPECT imaging offers increased sensitivity yet does not result in visible axial distortions generally present during half-cone-beam circular orbit acquisitions. Simulation studies of a disk phantom scan demonstrated that HCB circle-and-helix acquisition with the circular orbit

scan twice as long as the helical provided a 33% increase in efficiency over HCB helical-path, a 26% increase over fan-beam circular-orbit, and a 128% increase over parallel-hole beam circular-orbit acquisitions having the same spatial resolutions. Experimental Hoffman 3D brain phantom scans also demonstrated improved efficiency with HCB circle-and-helix acquisition. As a result, the reconstructed images displayed reduced noise and higher quality. Quantitative evaluation of detectability of simulated hot and cold lesions (rods) showed larger signal-to-noise ratios with HCB collimation compared to FB and PB, especially in the cranial part of the disk phantom. Triple-camera HCB circle-and-helix acquisition has clear advantages in brain SPECT imaging and offers potential improvement over conventional parallel-hole and fan-beam acquisitions.

ACKNOWLEDGMENT

We thank the anonymous reviewers for their insightful comments and constructive suggestions. One of the authors (RJ) is a consultant and officer of Data Spectrum Corporation (DSC) and has an equity interest in DSC. Another author (KLG) has served as a contractor to DSC.

This work was supported by the National Institute of Neurological Disorders and Stroke of the National Institutes of Health (NIH) under Grant Number R01 NS054797. Experimental data were acquired using shared instrumentation supported by the NIH National Center for Research Resources under Grant Number S10 RR15697.

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Fig. 1.

A gamma camera utilizing an HCB collimator positioned close to the patient's head. The HCB field of view is illustrated. The transaxial focal plane of the HCB collimator is perpendicular to the collimator surface.

Fig. 2.

A hot disk ($R=9$ cm) of the phantom containing 1 hot rod in the center ($r_0=0.72$ cm) and 4 cold rods ($r_1=0.36$ cm, $r_2=0.72$ cm, $r_3=1.08$ cm, $r_4=1.44$ cm) located 6.84 cm away from the center of the disk and separated by 90° . The hot rod was 50% hotter than the disk.

Fig. 3.

Circle-and-helix trajectories of HCB collimator focal points in a triple-camera SPECT system. Helical path is a combination of camera rotation and phantom translation in the caudal direction. The disk phantom consists of hot (grey) and cold (white) disks perpendicular to and centered on the AOR.

Fig. 4. Photon detection efficiencies per camera for (a)-(h) acquisition types described in section II.A without (top plot) and with (bottom plot) scatter and attenuation effects. Efficiencies are plotted for five hot disks of the phantom. Disks are numbered in a caudal-to-cranial order.

Fig. 5.

Reconstructed and post-filtered images of the five hot disks in the phantom: (a), (b) and (c) are the centrally located transaxial slices of the five hot disks for HCB circle-and-helix, FB and PB CO scans correspondingly. Profiles are drawn over rods r_0 , r_1 and r_3 . Figs. (d) and (e) are the central coronal and sagittal slices of the disk phantom shown for HCB, FB and PB acquisitions. Disks in transaxial, coronal and sagittal views are shown in a caudal-to-cranial order.

Fig. 6.

Reconstructed and post-filtered images of the Hoffman 3D brain phantom from HCB circle-and-helix (CH), FB and PB CO acquisitions along with the reference bitmap images. Three transaxial slices in a caudal-to-cranial order, and central sagittal (bottom left) and coronal (bottom right) slices are shown for each acquisition type and the bitmaps. Profiles are drawn over the images to compare the noise level and reconstruction quality.

TABLE I

Data Acquisition Types and Scan Trajectories

	Acquisition types	circular views	helical views	degrees /view	axial step/view
a)	HCB circular (CO)	360	0	1°	0 cm
b)	HCB multiple (6) CO	60 (×6)	0	2°	0 cm
c)	HCB circle-and-helix	240	120	1°c+ 2°h	0.108 cm
d)	HCB circle-and-helix	180	180	1°c+ 2°h	0.072 cm
e)	HCB circle-and-helix	120	240	1°c+ 2°h	0.054 cm
f)	HCB helical	0	360	1°	0.043 cm
g)	FB circular	360	0	1°	0 cm
h)	PB circular	360	0	1°	0 cm

Numbers of views per camera are shown. Equal time per view and equal total scan time is assumed. The notation '1°c+2°h' denotes 1° per circular (transaxial) view and 2° per helical (axial) view. For acquisition type (b) axial separation between the circular orbits was 2.2 cm.

TABLE II

Collimator Specifications

Coll. type	Focal length ^a (cm)	Hex hole size ^b (cm)	Septal thickness (cm)	Hole length (cm)	FWHM ^c (cm)	Point source sensit. ^d
PB	Inf.	0.138	0.016	4.54	0.65	4.44
FB	39	0.122	0.015	4.13	0.56	6.90
HCB	50	0.145	0.022	4.00	0.66	12.10

^aMeasured to patient side of collimator.

^bMeasured from the flat-to-flat surfaces of the hexagonal hole.

^cFull-width-at-half-maximum measured in air, and in transaxial focal plane at 12.5 cm from patient side of collimator.

^dSensitivity ((counts/s)/μ Ci) for a triple-camera SPECT system measured in air, and in transaxial focal plane at 12.5 cm from patient side of collimator.

TABLE III

Average Efficiencies for HCB, FB and PB Acquisitions

Acquisition type	Efficiency ($\times 10^{-4}$) w/o atten. & scatter	Efficiency ($\times 10^{-4}$) with atten. & scatter
(a)	1.58 ± 0.02	0.86 ± 0.01
(b)	0.98 ± 0.01	0.53 ± 0.01
(c)	1.33 ± 0.01	0.73 ± 0.01
(d)	1.21 ± 0.01	0.66 ± 0.01
(e)	1.08 ± 0.01	0.59 ± 0.01
(f)	0.99 ± 0.01	0.55 ± 0.01
(g)	1.14 ± 0.01	0.58 ± 0.01
(h)	0.64 ± 0.01	0.32 ± 0.01

Acquisition types are found in section II.A. Efficiencies (per camera) are averaged over five hot disks in the phantom.

TABLE IV

RMSE (%) of Reconstructed Hot and Cold Disks

Disk	RMSE (%) for acquisition types							
	(a)	(b)	(c)	(d)	(e)	(f)	(g)	(h)
Hot 1	5.0	5.2	3.8	3.9	3.8	3.8	3.9	5.1
Hot 2	15.2	3.5	3.8	3.4	3.3	3.4	3.9	5.1
Hot 3	29.0	5.0	3.5	3.1	3.0	3.1	3.9	5.1
Hot 4	26.6	6.6	3.7	3.1	2.9	3.0	3.9	5.1
Hot 5	22.2	2.1	2.9	2.2	2.0	2.1	3.9	5.1
Cold 1	2.4	0.4	0.3	0.1	0.0	0.0	0.0	0.0
Cold 2	6.1	0.8	0.5	0.1	0.0	0.0	0.0	0.0
Cold 3	16.0	0.2	0.5	0.2	0.1	0.0	0.0	0.0
Cold 4	23.8	0.7	1.1	0.5	0.2	0.1	0.0	0.0

RMSE of the reconstructed disks in terms of percentages of the true mean values of the hot disk activity. Acquisition types are found in section II.A. Disks are numbered in a caudal-to-cranial order.

TABLE V

Average Signal-to-Noise Ratio for Hot and Cold Rods

Disk number	Hot Rod r_0			Cold Rod r_1			Cold Rod r_2		
	HCB	FB	PB	HCB	FB	PB	HCB	FB	PB
1	3.0±0.1	2.7±0.2	2.2±0.1	3.4±0.2	3.4±0.3	2.5±0.2	8.1±0.3	8.8±0.3	6.0±0.2
2	3.2±0.1	3.1±0.2	1.9±0.2	4.4±0.3	3.4±0.2	2.7±0.2	9.3±0.2	8.0±0.3	6.3±0.2
3	3.7±0.1	2.9±0.2	2.0±0.1	4.5±0.3	3.7±0.2	2.8±0.2	9.6±0.2	8.2±0.3	6.2±0.2
4	3.8±0.2	3.0±0.1	2.3±0.2	4.6±0.2	4.1±0.3	2.5±0.2	10.2±0.4	9.2±0.3	5.7±0.2
5	7.3±0.3	2.9±0.2	2.2±0.1	7.6±0.4	3.6±0.3	2.4±0.2	15.7±0.5	8.9±0.2	6.3±0.2

Average SNR values for the central hot rod r_0 and the smallest cold rods r_1 and r_2 in the five hot disks, calculated using an ensemble of 16 noisy realizations. The disks are numbered in a caudal-to-cranial order. The uncertainties are calculated using Equation 4, assuming Gaussian distribution of SNR values.

TABLE VI**DETECTED NUMBER OF PHOTONS**

	HCB CH ($\times 10^6$)	FB CO ($\times 10^6$)	PB CO ($\times 10^6$)
Camera 1	4.45	3.27	2.31
Camera 2	4.47	3.24	2.32
Camera 3	4.47	3.25	2.33
Total	13.39	9.76	6.93

Number of photons detected by each γ -camera during HCB circle-and-helix, FB and PB circular-orbit acquisitions for equal scan time (adjusted for Tc-99m decay).