DUPLICATE

No. 779

A PROPOSAL OF DUAL FANBEAM GAMMA-RAY TRANSMISSION DATA ACQUISITION FOR QUANTITATIVE SPECT. **K. Ogawa and K. Yamamoto, Hosei Univ., Tokyo, Japan**

The purpose of the study is to reconstruct quantitative SPECT images. In order to reconstruct the quantitative SPECT images, it is necessary to correct attenuation of the gamma rays in the body, therefore, it is desired to obtain a true attenuation map measured by gamma-ray transmission CT. Many methods and geometries of the gamma-ray transmission data acquisition system have been proposed, however, there is a limitation in the size of the scintillator and this causes a serious artifact in reconstructed images of thorax. We propose a new geometry to acquire gamma-ray transmission CT data using fanbeam collimators. In the proposed data acquisition system, we use two detectors for the acquisition of gamma-ray transmission data. The two detectors equipped with the fanbeam collimator of the same focal length rotate together around the patient body. An external gamma-ray line source located at the focal point of the fanbeam collimators rotates simultaneously as the rotation of the detectors. From the two planar images (projection data) acquired by two detectors, we make an ideal transmission data having large field of view. The size of the ideal field of view can be controlled by the focal length of the fanbeam collimators. If the detector length is 40cm and the focal length changes from 50 cm to 60 cm, the radius of the effective field of view increases from 13.5 cm to 14.4 cm using a single detector. On the other hand, if we use the proposed data acquisition geometry, the radius increases from 19.2 cm to 23.5 cm, respectively. From the basic results, we made some simulations by changing the focal length and the detector size, and confirmed the effectivity of the proposed method. In summary, the proposed method could enlarge the field of view using an ordinary scintiallor with small size significantly, and the truncation of projection data was eliminated perfectly.

No. 780

FOCAL I-131 SPECT QUANTIFICATION BASED ON ACQUISITION OF MULTI-ENERGY-WINDOW DATA. **K.F. Koral, S. Lin, S. Buchbinder, and N.H. Clinthorne. University of Michigan Medical Center, Ann Arbor, MI.**

Accurate quantification of I-131 activity requires compensation for Compton scattering of gamma rays by the patient. With Tc-99m, dual-energy-window compensation has been shown to be complicated by the uncertainty of the background level (Luo et al, Phys Med Biol 1995; 40(1)). For I-131, it is further complicated by collimator scattering and by septal penetration of high-energy (up to 723 keV) gamma-ray emissions. We here propose a solution based on acquisition of multi-energy-window data which should soon be possible with at least one commercial system using list mode. Our solution requires processing local energy spectra with deconvolution fitting [IEEE TMI 1992; 11:351-360]. It replaces the previously-utilized "scatter-free" spectrum with a "quantification" spectrum which is to contain only gammas which have not scattered in the patient or the collimator and which deposit their full energy in the crystal; high-energy emissions are thus elliminated.

To test the method we acquired multi-energy-window data for an elliptical water phantom filled with 4 levels of I-131 background (including none). The ellipse also contained a hot, 6-cm-diameter, off-axis sphere. A special MAC II-based system was connected to a standard GE400AT gamma camera to achieve the x-y-E imaging. The quantification energy spectrum was approximated by imaging a small, thin source and using only the counts within a 4-pixel region of interest centered on the point-source image. Reconstruction was by the EM-ML algorithm. The conversion constant from reconstructed counts to activity was set by the zero-background experiment.

Within a 20% window, the measured quantification spectrum contains only 16.3% of the counts present for the entire camera face; it sits on a much-lower-strength, sloping-straight-line pedestal. Quality of the quantification method is assessed by the accuracy of the estimate of sphere activity in the three cases of non-zero background level.

No. 781

IMPLEMENTATION OF A RECURSIVE TWO-DIMENSIONAL MAXIMUM LIKELIHOOD POSITION ESTIMATOR FOR SCINTILLATION CAMERAS. C.Y. Ng, N.H. Clinthorne, W.L. Rogers. Division of Nuclear Medicine, the University of Michigan, Ann Arbor, MI.

The most widely used position estimation method for scintillation cameras is the centroid estimator which is basically a calculation of the distance weighted mean. This approach is simple but has the drawback that it disregards the statistical nature of photon interaction and detection. The maximum likelihood estimator is superior in this aspect. Clinthorne and Rogers et.al. implemented a maximum likelihood position estimator for SPECT. They assume that the (x,y) coordinates can be estimated independently in order to save computer memory. However, this is valid only when the sampled light-spread function (LSF) is separable and will lead to spatial distortion when the constraint is violated. This led Liu and Clinthorne et.al. to investigate the possibility of using a recursive calculation for the two dimensional maximum likelihood position estimate. By using computer simulation, they demonstrated that recursive 1-dimensional estimates of the most likely 2-d location of a scintillation event converges rapidly to within a fraction of a millimeter of the true location. In this work, the recursive 2-d estimation is implemented on a modular scintillation camera using software for position estimation. The LSF for x and y were determined by translating a highly collimated line source in 0.5 or 1 mm increments. LSF were determined on a grid of 5 mm and linearly interpolated to 1 mm grids. Weights for position estimation were generated according to these LSFs. Performances of the independent 2-d estimator and recursive 2-d estimator were then compared by estimating the position of a point source which was translated in the x direction. In less than 6 iterations, the recursive 2-d estimator gave an estimate which is within 3 mm of the true position, while the independent 2-d estimator gave a result which could deviate as much as 17 mm from the true position for this highly distorted region of the detector. Further improvement would be expected by using a point source for LSF measurement and bilinear interpolation to a finer grid.

Thus the recursive 2-d estimator gives a low distortion estimate of the location of scintillation event with much improved linearity.

No. 782

A HYBRID-GRID PARAMETERIZATION METHOD FOR SPECT RECONSTRUCTION Y. Zhang, J. A. Fessler, N. H. Clinthorne, W. L. Rogers, Bioengineering Program, The University of Michigan, Ann Arbor, MI.

To incorporate structurally correlated high resolution MRI anatomical structures into SPECT reconstruction to improve quantitative results, SPECT and MRI pixel sizes must first be matched. MRI pixels are much smaller (usually 0.2mm) than those of SPECT because of their higher resolution. If we use a finer SPECT pixel grid to match the MRI pixel size, we will have a larger parameter set and a larger ECT system response matrix. Consequently, image reconstruction will require larger computer memory and much greater computing effort, which greatly increases the reconstruction time.

We have investigated an approach to circumvent the huge system and image matrices due to pixel matching while maintaining accuracy within a region of interest(ROI). In this approach, the SPECT image domain is parameterized into a hybrid grid structure, only the ROI is parameterized in terms of the MRI resolution, while the rest of the image pixels can still be the original SPECT pixel size or even coarser. The transition probabilities are also generated according to the hybrid pixel grid. An FBP result can be used to select the ROI and initialize the hybrid grid image for further iterative reconstruction with the corresponding hybrid weights. At this stage, the structual information from MRI can be readily incorporated into the fine pixel region of the hybrid SPECT image using different methods. If an ROI is $1/\alpha$ of the whole image, and the ratio of MRI and SPECT pixels is β, then the ratio of the number of hybrid pixels to the number of fine-grid-only pixels is $((\alpha-1)/\alpha+\beta^2/\alpha)/\beta^2$, i.e. when $\alpha=4$ and $\beta=8$, which is