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Estimation of 6-Degree-of-Freedom (6-DOF) Rigid-Body Patient Motion From Projection Data by the Principal-Axes Method in Iterative Reconstruction

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Abstract

We developed a unique method for estimating and compensating rigid-body translations and rotations from scatter and-attenuation-compensated projection data in iterative reconstruction when multiple projection angles are acquired at the same time. During reconstruction, both the non-attenuated and attenuated line-integrals are calculated. Their ratios are then multiplied to the scatter-corrected projection data to estimate scatter-and-attenuation-compensated projection data. At the end of each iteration, the sets of compensated projection data for the angles acquired at the same time are employed to calculate the center-of mass and the inertia tensor, which are used to estimate the location and orientation of the imaging object by the principle-axes method. The estimated motion is applied in the next iteration to reposition the estimated slices and attenuation map in the projector and back-projector to match the pose of the patient at time the projections were acquired. To evaluate our method, we simulated an acquisition of the MCAT phantom with a 3-head SPECT system and imaged the Data Spectrum anthropomorphic phantom on a 3-head IRIX SPECT system. In simulations the phantom translated and rotated by the same amount 9 times. A numerical projector modeling the motion, attenuation, and distance-dependent blurring was used to generate the projection data. Poisson noise was added and 30 noise-realizations were generated. In the experiment with the anthropomorphic phantom, four 360-degree acquisitions were performed with the phantom translated or rotated beforehand. A motion-present dataset was made by mixing the 4 acquisitions. For both the MCAT phantom simulations and anthropomorphic phantom experiment, the motion-present data were reconstructed with 10 iterations of the OSEM which estimates and corrects the motion as described above. Our method obtained visually artifact-free reconstructions, while the reconstruction with no motion correction showed severe artifacts. The motion estimated from our method was in good agreement with the motion simulated. We determined in MCAT simulated and actual phantom acquisitions that our data-driven approach was effective reducing motion artifacts.

Index Terms

Iterative reconstruction; rigid-body motion; single photon emission computed tomography

I. INTRODUCTION

IN SPECT and PET, patient motion during imaging can cause reconstruction artifacts and degradation of diagnostic accuracy if not properly compensated. One major difference between SPECT and PET when it comes to motion correction is that with PET all angles are generally acquired all the time; whereas, in SPECT only 2 to 3 angles (depending on the number of camera heads) are acquired at the same time. For SPECT some approaches correct for only axial and lateral translational motion in the sinogram domain [1]–[4], other methods aim to estimate the six-degree-of-freedom (6-DOF) transformation of the activity distribution in three-dimensions (3D) [5]–[7]. In reality patient motion is generally non-rigid in nature, except in cases imaging hard objects, e.g., head imaging. Correction of non-rigid motion is more difficult since it involves more degrees-of-freedom than correction of rigid-body motion. In presence of non-rigid motion, rigid-body motion correction techniques might still be used to correct for the rigid-body component of the general non-rigid motion.

In this work we present a unique approach for using the 2D projection data in SPECT/PET to calculate the inertia tensor used for 6-DOF rigid-body motion detection and correction. Our data-driven method differs from the other methods in that it estimates the inertia tensor from scatter-and-attenuation-compensated projection data, and then calculates analytically the orientation and translation of the patient by the principle-axes method [8]. It requires that for each motion state there are at least projections for three different angles available. This is guaranteed for a PET or 3-head SPECT system such as the older Philips Prism3000's and Irix, and the new Digirad Cardius. For infrequent patient motion, it is also possible to apply our method to a dual-head or even single-head SPECT camera by grouping multiple acquisition angles into one motion state.

In this work we assumed that no patient motion occurs during acquisition of each view for the reason of simplicity (patient moves only during camera rotation). Our method could be applied in principle to the general case by re-histogramming the list-mode data to subdivide each projection into many sub-projections which can be approximated by single motion states. In this work we concentrated on the theoretical part of the work and left alone the detailed practical considerations as the future work.

II. METHODS

We will present our unique method of correction for the rigid-body motion in Section II-A. Our method only applies to the parallel-beam geometry. It consists of two aspects: 1) determination of the 6-DOF motion from the ideal projection data, and 2) iterative estimation of the motion for SPECT data from the scatter-and-attenuation-compensated projection data. To evaluate our method, we will present simulations in Section II-B and phantom experiments in Section II-C.

A. Detection of 6-DOF Rigid-Body Motion From Scatter-and-Attenuation-Compensated Projection Data in the OSEM Reconstruction

From the classical theory of the rigid-body motion, the principle- axes method has been developed and applied to register 3D tomographic images. The principle-axes method requires calculation of the inertia tensor

$$I_{ij} = \iiint (r^2 \delta_{ij} - r_i r_j) \rho dv, i, j=1, 2, 3, \quad (1)$$

where ρ is the radioactive tracer distribution, dv is an infinitesimal volume element, $\mathbf{r}_1, \mathbf{r}_2, \mathbf{r}_3$ and $\mathbf{r}$ are the x, y, z displacement and the distance from the center-of-mass point $\mathbf{r}_1^C, \mathbf{r}_2^C, \mathbf{r}_3^C$, respectively, $\delta_{ij} = 1$ for $i = j$, and zero otherwise. To calculate I_{ij} , we need only to know

$$T_{ij} = \iiint r_i r_j \rho dv, i, j=1, 2, 3, \quad (2)$$

which is a symmetric tensor that has 6 independent components. It is also straightforward to show that $\mathbf{T}$ has the same eigenvectors as $\mathbf{I}$. Actually we use $\mathbf{T}$ instead of $\mathbf{I}$ to calculate the principal-axes which define the orientation of the object and thereafter when we mention inertia tensor we mean $\mathbf{T}$.

In the “prime” coordinate system rotating with the camera (Fig. 1), at camera angle α , the tensor $\mathbf{T}$ is transformed to

$$T'_{ij}(\alpha) = \iiint r'_i r'_j \rho dv = \sum_{k=1}^3 \sum_{l=1}^3 M_{ik}(\alpha) T_{kl} M_{jl}(\alpha), \quad (3)$$

where $\mathbf{M}(\alpha)$ is the transformation matrix which maps a point in the global coordinate system to the rotating coordinate system. Written explicitly,

$$\mathbf{M}(\alpha) = \begin{bmatrix} \cos\alpha & \sin\alpha & 0 \\ -\sin\alpha & \cos\alpha & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad (4)$$

First, if we assume there is no attenuation, scatter or other degradations present in the projection data for a parallel-beam system, for a quantity H that does not vary along the collimation (y') direction, we have

$$\iiint H \rho dv = \iiint H \rho dl da = \iint H da (\int \rho dl) = \iint H p da \quad (5)$$

where the projection data p equal to the line integral $\int \rho dl$, da is an infinitesimal area element on the camera plane.

We can estimate the following quantities from the planar integral of projection data p_α :

$$T'_{11}(\alpha) = \iint (r'_1)^2 p_\alpha da, \quad (6)$$

$$T'_{33}(\alpha) = \iint (r'_3)^2 p_\alpha da, \quad (7)$$

$$T'_{13}(\alpha) = T'_{31}(\alpha) = \iint r'_1 r'_3 p_\alpha da, \quad (8)$$

$$(r_1^C(\alpha))' = \frac{\iint x' p_\alpha da}{\iint p_\alpha da}, \quad (9)$$

$$\text{and } (r_3^C(\alpha))' = \frac{\iint z' p_\alpha da}{\iint p_\alpha da}, \quad (10)$$

where $r'_1 = x' - (r_1^C)'$, and $r'_3 = z' - (r_3^C)'$.

For quantities that involve y' , such as $T'_{12}(\alpha)$, $T'_{22}(\alpha)$, $T'_{23}(\alpha)$, and $(r_2^C(\alpha))'$, the volume integrals cannot be replaced by planar integrals, since y' changes along the line integral direction in (5).

Writing (3) in terms of components, we have

$$T'_{11}(\alpha) = T_{11} \cos^2 \alpha + 2T_{12} \sin \alpha \cos \alpha + T_{22} \sin^2 \alpha, \quad (11)$$

$$T'_{13}(\alpha) = T_{13} \cos \alpha + T_{23} \sin \alpha, \quad (12)$$

$$T'_{33}(\alpha) = T_{33}. \quad (13)$$

If for one motion state there are at least three frames (e.g., $\alpha_1, \alpha_2, \alpha_3$, and $\alpha_3 > \alpha_2 > \alpha_1$) of projections available, the tensor $\mathbf{T}$ can be solved from (11)–(13). In case that we have three angular views for a motion state, we can rewrite (11), (12) as

$$\begin{bmatrix} \cos^2 \alpha_1 & 2\sin \alpha_1 \cos \alpha_1 & 0 & \sin^2 \alpha_1 & 0 \\ 0 & 0 & \cos \alpha_1 & 0 & \sin \alpha_1 \\ \cos^2 \alpha_2 & 2\sin \alpha_2 \cos \alpha_2 & 0 & \sin^2 \alpha_2 & 0 \\ 0 & 0 & \cos \alpha_2 & 0 & \sin \alpha_2 \\ \cos^2 \alpha_3 & 2\sin \alpha_3 \cos \alpha_3 & 0 & \sin^2 \alpha_3 & 0 \\ 0 & 0 & \cos \alpha_3 & 0 & \sin \alpha_3 \end{bmatrix} \begin{bmatrix} T_{11} \\ T_{12} \\ T_{13} \\ T_{22} \\ T_{23} \end{bmatrix} = \begin{bmatrix} T'_{11}(\alpha_1) \\ T'_{13}(\alpha_1) \\ T'_{11}(\alpha_2) \\ T'_{13}(\alpha_2) \\ T'_{11}(\alpha_3) \\ T'_{13}(\alpha_3) \end{bmatrix}. \quad (14)$$

The matrix (14) can be solved in the least-squares sense, or analytically by using only five of six equations. Note that if $\alpha_2 = \alpha_1 + \pi$ or $\alpha_3 = \alpha_1 + \pi$ or $\alpha_3 = \alpha_2 + \pi$, (14) becomes ill-posed and has multiple solutions.

Similarly, the center-of-mass can be solved from (9, 10) and the transformation rule for vectors if at least two views (e.g., α_1 , α_2 , and $\alpha_2 > \alpha_1$) are available for a motion state, either analytically or in the least-squares sense. In the case of two views, we have

$$\begin{bmatrix} r_1^C \\ r_2^C \\ r_3^C \end{bmatrix} = \begin{bmatrix} \cos\alpha_1 & \sin\alpha_1 & 0 \\ \cos\alpha_2 & \sin\alpha_2 & 0 \\ 0 & 0 & 1 \end{bmatrix}^{-1} \begin{bmatrix} (r_1^C(\alpha_1))' \\ (r_1^C(\alpha_2))' \\ \frac{(r_3^C(\alpha_1))' + (r_3^C(\alpha_2))'}{2} \end{bmatrix}. \quad (15)$$

Equation (14) and (15) are the main theoretical results of this work which describes how to solve the Inertia tensor and the center-of-mass from 2D projection data. Combined with the principal-axes method, as shown afterwards, 6-DOF rigid-body motion can be determined from these two equations.

To summarize, we need at least three simultaneous views to determine the tensor $\mathbf{T}$ and the center-of-mass (r_1^C, r_2^C, r_3^C) . This condition is guaranteed for a PET or a 3-head SPECT system (with no cameras parallel to each other). Once $\mathbf{T}$ is known, the principal-axes method calculates the eigenvectors of $\mathbf{T}$, and the rotation matrix from two sets of eigenvectors ($\mathbf{v}_1, \mathbf{v}_2$, and $\mathbf{v}_3$ at time t , $\mathbf{v}'_1, \mathbf{v}'_2$, and $\mathbf{v}'_3$ at time t') for $\mathbf{T}$ at two different time points [9]. Since the eigenvector rotates with the object, $\mathbf{v}'_1, \mathbf{v}'_2$, and $\mathbf{v}'_3$ and are related to $\mathbf{v}_1, \mathbf{v}_2$, and $\mathbf{v}_3$ by rotation matrix $\mathbf{M}$ as

$$\mathbf{v}'_1 = \mathbf{M}\mathbf{v}_1, \mathbf{v}'_2 = \mathbf{M}\mathbf{v}_2, \mathbf{v}'_3 = \mathbf{M}\mathbf{v}_3. \quad (16)$$

Since $\mathbf{T}$ is symmetric, its eigenvectors are perpendicular to each other. Therefore, we can construct $\mathbf{M}$ as

$$\mathbf{M} = \mathbf{v}'_1 \otimes \mathbf{v}_1 + \mathbf{v}'_2 \otimes \mathbf{v}_2 + \mathbf{v}'_3 \otimes \mathbf{v}_3 \quad (17)$$

where $\otimes$ denotes the tensor product of two vectors.

The translations can be calculated from the movement of the center-of-mass.

For ideal data consistent with the line-integral model the 6-DOF rigid-body motion can be exactly determined by (15)–(17). In reality, attenuation, scatter, and noise will modify the projection data such that the sensed principal axes will not be consistent with angle. For PET pre-correction of the projections for attenuation and scatter can be applied. For SPECT, we estimate iteratively the motion from “scatter-and-attenuation-compensated” projection data, which are estimated iteratively as part of reconstruction. Scatter can be estimated by the ESSE [10], ESSI [11] or TEW [12] methods, and subtracted from the projection data. During reconstruction, both non-attenuated and attenuated line-integrals are calculated and their ratios are multiplied to the scatter-subtracted projection data to obtain the scatter-and-attenuation-compensated projection data (Fig. 2). At the end of each OSEM iteration, the

motion relative to the reference view (view # 0) is solved from the scatter-and-attenuation-compensated projection data for all views (the simultaneous views have the same motion) and applied to the next iteration in the way we reconstruct with “known rigid-body motion” [13]. The OSEM subsets are divided in the regular way and not restricted to grouping simultaneous views into one subset. Note that the image is reconstructed from the measured projection data with compensation for attenuation, scatter, and detector resolution, not from the scatter-and-attenuation-compensated projection data.

B. The MCAT Phantom Simulations With a 3-Head SPECT System

To evaluate our method, we simulated acquisitions of the MCAT phantom with a 3-head SPECT system (120 degrees apart among heads). Each head rotated 120 degrees by 40 steps for a total acquisition of 360 degrees. The phantom translated and rotated by the same amount 9 times, which resulted in 10 motion-states for the phantom. The overall motion was 2, 3, and 2.5 cm translations along x, y, z directions, and a 15-degree rotation by the vertical-axis. A numerical projector modeling the motion, attenuation, and distance-dependent blurring (with parameters derived for Tc-99m) was used to generate the projection data. And no scatter was included in the simulations. The counts in the heart was scaled to 0.5 million, which resulted in 2 million counts in the liver. Poisson noise was added at this level and 30 noise realizations were generated. The motion-present data were reconstructed with 10 iterations of the OSEM (4 subsets, 30 views per subset) which estimates and corrects the motion as described above. Note that division of the OSEM subsets considers no motion states of the views. At end of each OSEM iteration the motion for each simultaneous views (contains 3 views here) was estimated and used for the next iteration. The motion estimated was output at the end of reconstruction. For comparison, the same data were also reconstructed with compensation for the true motion and without motion compensation.

C. Spect Acquisitions of the Anthropomorphic Phantom

Tc-99m was added to the Data Spectrum anthropomorphic phantom such that there was an equal concentration in the heart and liver, and the background was at 10% of their concentration. The phantom was then imaged on an IRIX 3-head SPECT system. Four 360-degree emission acquisitions were performed with photopeak (centered on 140.5 keV with 15% width) and TEW scatter (centered on 120 keV with 5% width) windows. Between acquisitions the phantom was translated or rotated. The orientation and location of the phantom were tracked by a visual tracking system (VTS) [14], [15]. A motion-present dataset was obtained by mixing the 4 acquisitions. Beacon transmission imaging [16] was performed when the phantom was in the baseline configuration. The motion-present dataset was reconstructed by 10 iterations of the OSEM (4 subsets, 30 views per subset) that incorporates our motion estimation and correction method, and reconstructed without motion correction. For comparison, the motion-free dataset at baseline (before motion was applied) was also reconstructed. In all these reconstructions, attenuation compensation and detector resolution compensation were applied [17].

III. RESULTS

A. MCAT Phantom Simulations With a 3-Head SPECT System

In the average of reconstruction of the 30 noisy motion-present datasets, our method obtained a visually artifact-free reconstruction, almost identical to the reconstruction with the true motion, while the average reconstruction with no motion correction shows severe artifacts (Fig. 3). Reconstruction shows fast convergence, as illustrated in the plots of the log-likelihood versus the OSEM iteration number (to obtain more data points for the plot, reconstruction was performed by 50 iterations of the OSEM with 4 subsets and 30 angles per subset) (Fig. 4). The motion estimated from our method is in good agreement with the motion simulated (Figs. 5–8). In Fig. 5, the estimated translations averaged over 30 noise-realizations were plotted along with the true translations, which matched with each other very well at all angles, with the maximum error of 0.08 cm that was much smaller than the 2 cm system resolution. The error bars for the estimated translations were plotted in Fig. 7, with the maximum standard deviation of 0.21 cm. The three Euler angles (also called Tait-Bryan angles, with X-Y-Z conventions) were estimated to represent the sequential rotations along X, Y, and Z axes and were plotted in Figs. 6 and 8, which show the maximum error of 1.4 deg and the maximum standard deviation of 1.8 deg. The motion detection and correction slows the reconstruction by less than 10%.

B. SPECT Acquisitions of the Anthropomorphic Phantom

The motion-present dataset was reconstructed without motion correction and with motion correction by our method. For comparison, a motion-free dataset was also reconstructed. In the case of motion correction, two options were tested for the scatter: (1) no scatter correction, and (2) scatter correction with TEW using the two-window version of TEW for Tc-99m. A transaxial slice for each case is shown in Fig. 9. The motion correction with our method greatly reduced the motion artifacts. However, scatter compensation with TEW seemed to worsen the motion estimation. This is probably because TEW is an approximate scatter correction method, and introduces additional noise to the corrected projection data though the relatively low-count scatter window projection data. The motions estimated from our data-driven approach were compared with the VTS results in Table I.

IV. DISCUSSIONS

Our method requires the estimation of the ideal Radon transform from the attenuation and scatter corrupted projection data. Since both attenuation and scatter depends on the source distribution, an iterative estimation approach is adopted. The center-of-mass and tensor $\mathbf{T}$ (a surrogate of the Inertia tensor) should therefore be calculated from attenuation-and-scatter-compensated projection data. For the experiment with the anthropomorphic phantom filled with Tc-99m, however, we found that a simple scatter correction method worsened the motion estimation. Thus care must be applied when performing correction. As future work, we would like to investigate the influence of the more sophisticated methods like ESSE [10] and ESSI [11] on the motion detection for our approach.

There is no theoretical proof that the iterative motion estimation will converge by our method, though it converged practically for all the data in this work. In cases when the motion estimation is not converging, our method could be modified by applying relaxation to the motion estimation (apply fractional increments at each step) and/or changing the motion updating frequency (e.g., update once per two iterations), which may help stabilize the motion estimation.

Our approach is developed solely for parallel-beam geometry. Extension to other imaging geometries such as fan-beam, cone-beam may not be possible, since (5) is no longer valid in non-parallel-beam cases.

In the case that different organs move differently, regions-of-interest (ROIs) might be applied [18]. The projection data could be separated iteratively for each ROI according to their contributions estimated from the projector, and the motion could be estimated for each ROI.

As a data driven method, our approach may fail to work if there is severe data truncation due to very large motion or due to the small detectors. An exception could be when data truncation occurs only for an isolated organ which is positioned axially away from the major imaging target organ. In this case with application of the above ROI method our method may still work.

It is also possible to apply our method to SPECT systems with less than 3 heads, by grouping adjacent views together and assuming there is no severe motion during the acquisition of these views. A big concern for this approach is that solving for the motion might be sensitive for the noise and other degradations in the data. Further investigations are needed toward SPECT systems with one or two heads.

Our method may be applied to correct for the rigid-body type motion, such as for head imaging with PET or parallel-beam SPECT, and the rigid-body component of non-rigid motion, such as the respiratory motion in SPECT or PET cardiac studies. To successfully apply our approach to real patient data, we may need to constrain the motion (maximum translations or rotations) and/or reduce degrees-of-freedom (e.g., only translations allowed) in estimation. Applying relaxation to the estimated motion and changing the updating frequency for the motion are among possible techniques to stabilize the motion estimation for real world data.

V. CONCLUSION

As a data-driven approach, our method is shown to work effectively for a three-head SPECT system which employs parallel collimators.

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Fig. 1.

The global and rotating coordinate systems as used in derivation of formulas.

Fig. 2.

The flow chart of our reconstruction methodology. At the end of iteration, the inertia tensor and center-of-mass are solved for each view from the attenuation-and-scatter-corrected projection data, and the 6-DOF motion from the initial reference view is estimated by the principal-axes method and applied in the next iteration of reconstruction.

Fig. 3.

The same transaxial slice from reconstructions of the motion-present data. (Left) Motion correction with the true motion. (Middle) No motion correction. (Right) Motion correction with our data-driven method. In each case the image was an average over the 30 noise-realizations.

Fig. 4.

The plots of the log-likelihood vs. the OSEM iteration number in reconstruction of the motion-present data. Solid line—motion correction with the true motion. Dotted line—motion correction by our data-driven approach. Dashed line—no motion correction.

Fig. 5.

The translations were estimated by our method and averaged over the 30 noise-realizations, compared with the truth.

Fig. 6.

The Euler angles for the rotation were estimated by our method and averaged over the 30 noise-realizations, compared with the truth.

Fig. 7.

Plots of the average translations and error bars (SD) estimated from the 30 motion-present noise-realizations of the MCAT phantom simulations, versus the camera angle for head one of a three-head SPECT system. Each head rotates 120 degrees in acquisition.

Fig. 8.

Plots of the average Euler angles and error bars estimated from the 30 motion-present noise-realizations of the MCAT phantom simulations, versus the camera angle for head one of a three-head SPECT system. Each head rotates 120 degrees in acquisition.

Fig. 9.

(Upper Left) A transaxial slice from reconstruction of the motion-free acquisition of the Data Spectrum anthropomorphic phantom. (Upper Right) The same slice from the reconstruction of the motion-present acquisition without motion correction. (Lower Left) The same slice with motion correction by our method and with TEW for scatter correction in determination of motion. (Lower Right) The same slice with motion correction and no scatter correction. Note that the left and right edges for each image were truncated due to zooming used for display.

TABLE I

THE MOTION WAS ESTIMATED BY THE DATA-DRIVEN APPROACH WITH TEW (THE FIRST ROW), WITHOUT TEW (THE SECOND ROW), AND THE VTS (THE THIRD ROW). THE STANDARD DEVIATIONS WERE CALCULATED FROM EACH SUBGROUP (10 VIEWS) IN WHICH THE PHANTOM REMAINED STATIONARY. x, y, z,-TRANSLATIONS IN X, Y, AND Z. ROT X, ROT Y, ROT Z-EULER ANGLES AROUND X, Y, AND Z AXES

	Step 1	Step 2	Step 3	Step 4
x(cm)	0.40±0.24 0.27±0.17 0	0.89±0.06 0.70±0.32 0.07	0.80±0.14 0.72±0.32 0.29	0.81±0.08 0.68±0.11 0.47
y(cm)	0.09±0.11 0.01±0.05 0	0.29±0.10 0.38±0.14 0.06	3.03±0.08 2.72±0.06 2.11	0.98±0.11 0.79±0.08 0.21
z(cm)	0.09±0.11 0.02±0.03 0	-4.13±0.14 -3.84±0.02 -4.05	2.19±0.14 1.99±0.02 1.92	-3.06±0.23 -2.87±0.04 -3.24
Rot x	3.42±2.08 1.06±0.84 0	4.69±1.12 0.62±0.55 0.02	0.59±1.57 0.94±0.49 0.11	-0.75±1.04 0.66±0.90 -0.86
Rot y	0.39±0.40 0.05±0.21 0	0.98±0.48 0.18±0.30 -0.149	1.06±0.13 0.42±0.19 -0.185	4.02±0.26 3.52±0.17 3.72
Rot z	-0.39±3.70 0.36±0.71 0	0.89±1.94 0.33±0.39 0.37	-0.08±1.68 0.08±0.64 0.33	-10.00±0.79 -10.20±0.73 -8.79