Registration of Planar Emission Images with Reprojected CT Data

Helena Surová-Trojanová, W. Craig Barker, Jorge A. Carrasquillo, and Stephen L. Bacharach

Nuclear Medicine Department, Warren Grant Magnuson Clinical Center, National Institutes of Health, Bethesda, Maryland

Planar γ-camera imaging is still widely used clinically. Alignment of planar images with images from tomographic modalities, such as CT, or with other planar images would be desirable. Here, we present and evaluate a method for such an alignment, using planar transmission images acquired with the emission images and reprojection of the 3-dimensional CT data. This method permits determination of which CT slice corresponds to a particular row of pixels in the γ-camera image and which column of pixels in that CT slice corresponds to a particular pixel in the emission data. **Methods:** A method based on maximization of the correlation coefficient, previously used for 3-dimensional datasets, was modified to permit 2-dimensional registrations. Planar transmission measurements were obtained using a collimated ^{99m}Tc flood source in conjunction with planar emission studies. The CT data were first reprojected to permit the 2-dimensional registration. The registration method was evaluated for its accuracy and reproducibility. **Results:** For phantom data, the registration errors were −0.1 ± 1.0 mm for x-translations, 1.0 ± 1.3 mm for y-translations, and −0.2 ± 0.3° for rotations. For patient data, the errors were 1.6 ± 0.8 mm for x-translations, 1.3 ± 1.0 mm for y-translations, and 0.5 ± 0.5° for rotations. An examination of the need for rescaling of the attenuation data (to compensate for the different photon energies used in the respective attenuation measurements) showed no significant impact on registration error. When 5 different regions of interest were used for the correlation coefficient calculation, the mean errors attributable to region-of-interest choice alone were 1.0 mm for x-translations, 2.0 mm for y-translations, and 1.2° for rotations. **Conclusion:** In almost all instances, translational registration errors were kept to subpixel levels (pixel size, 2.6 mm) and rotational errors to 1° or less. The 1 exception was in the easily avoidable case of "pitch" rotations of the patient of 2° or more. The modified registration method provides a simple yet reliable way to provide cross-modality evaluation of planar emission data.

Key Words: planar radionuclide scintigraphy; CT; multimodality registration

J Nucl Med 2000; 41:700–705

Despite recent advances in tomographic scintigraphy, planar imaging remains a major component of clinical nuclear medicine practice. Combining data obtained from planar imaging with information from other imaging modalities, such as CT, is often desirable. This combination would be useful both to better define anatomy and to refine quantitative analyses, such as geometric mean imaging—a long-accepted method for quantitating data from planar images. For example, monoclonal antibody imaging has recently been used to obtain dosimetric data before therapeutic administration (*1,2*). Geometric mean planar imaging has been a key component of these dosimetric measurements. The traditional region-based geometric mean method can be extended on a pixel-by-pixel basis through the addition of planar transmission images (*3*). Because we already routinely acquire this transmission data at our institution, we investigated the feasibility of registering this planar 2-dimensional data with 3-dimensional CT data obtained concurrently to more accurately determine the corroborative information derived from the CT data.

Through the alignment of reprojected CT data (rCT) with planar emission images, one can potentially determine which CT slice corresponds to a particular row of interest in a related planar emission image (Fig. 1). The planar-to-rCT registration would also enable one to determine which row of pixels in the CT slice corresponds to a particular pixel in the emission image, thereby providing great potential benefit to dosimetry calculations or therapy considerations. Finally, planar–rCT alignment would allow one to overlay the functional planar emission data with the anatomic rCT.

A technique for the 3-dimensional registration of 2 cardiac PET attenuation scans of the same subject acquired at different times based on maximizing the pixel-to-pixel correlation coefficient has been described (*4*). By adapting this methodology to the 2-dimensional case, it should in principle be possible to align sequential planar transmission images with one another as well as with rCT. Because both planar transmission and CT datasets are derived from the attenuation properties of an object, such an adaptation seems feasible. Planar transmission images can be obtained with the patient positioned as during emission imaging by acquiring short-duration (2- to 8-min) transmission data immediately before or after the emission imaging. This practice is common in PET. Alternatively, where practical, both emission and transmission data can be acquired simultaneously. Alignment of rCT with planar transmission data can consequently provide a way to align rCT to planar emission images. In this study we test, using human and phantom studies, the accuracy, precision, and reproducibility

Received Oct. 25, 1998; revision accepted Aug. 31, 1999.

For correspondence or reprints contact: Craig Barker, PhD, Nuclear Medicine Department, National Institutes of Health, Bldg. 10, Rm. 1C-401, 10 Center Dr., MSC 1180, Bethesda, MD 20892-1180.

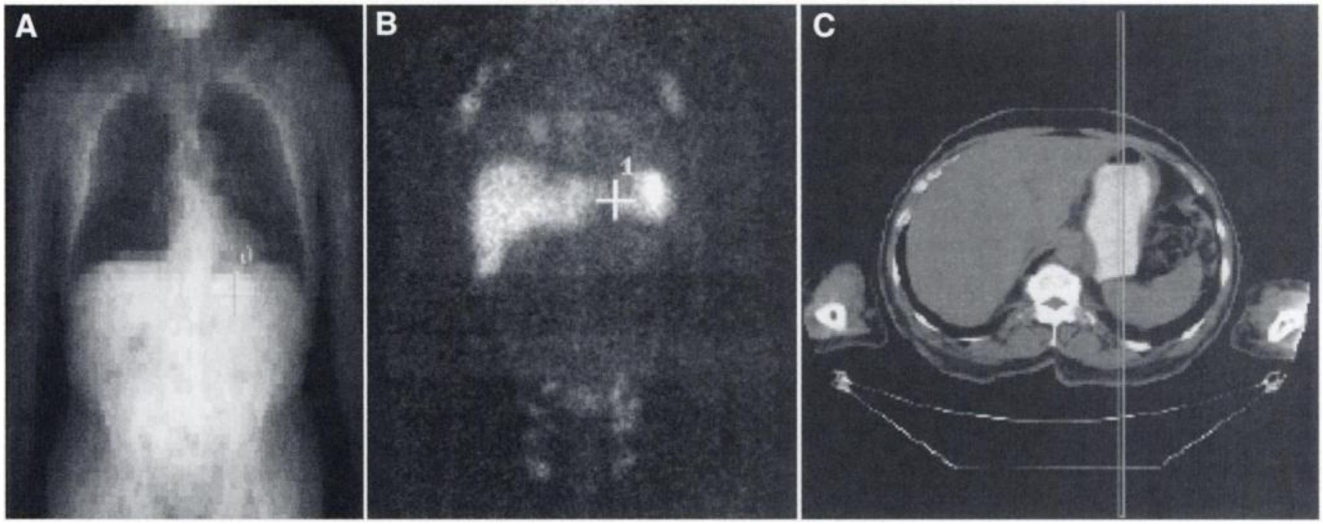

FIGURE 1. Simultaneous display of registered anatomic (CT) and functional (planar emission) data. After registration, user positions marker on planar emission scan (B). Corresponding position is indicated on rCT (A) and corresponding CT slice (C).

of this methodology to align rCT with planar transmission data.

MATERIALS AND METHODS

Registration Procedure

Many methodologies have been developed to register pairs of medical image sets (*5–9*). These methods often assume that the 2 datasets come from imaging modalities that derive their respective data from different physical properties of the object. In our application, both planar transmission and CT datasets are derived from the same physical effect, namely photon attenuation. Consequently, a method based on the maximization of the correlation coefficient between the 2 image sets is expected to be a simple and effective means to achieve accurate registration. In the 3-dimensional implementation of the correlation-based registration algorithm, a volume of interest that excluded the patient bed and included most of the lungs and some of the liver was used to determine which pixels to use for the correlation calculation (*4*). This study uses the 2-dimensional equivalent of this approach. We registered planar anterior chest views using the pixels within a region of interest (ROI) that always included the upper third of the liver and at least the lower half of the lung. The lateral external body surfaces were excluded from consideration, because they may deform excessively in different imaging situations (e.g., if the CT bed is curved but the planar table is flat). Indeed, the 3-dimensional implementation showed that the lung border regions (near the heart, liver, and inner chest wall) were primarily responsible for the high correlation of the registered images. The lung regions affect registration most because of the large attenuation differences between lung and other tissue. A sample ROI used with the 2-dimensional implementation is shown in Figure 2.

To achieve registration between 2-dimensional image datasets, 2 orthogonal translations (1 perpendicular to the long axis of the patient and 1 along the long axis, both in the plane of the image) followed by 1 rotation (around the axis perpendicular to the plane of the image) were performed iteratively until the maximum pixel-to-pixel correlation was found. Only rigid body translations and rotations were applied. In practice, we determined that the maximum could be reached in 8 iterations or less. Given this result, we limited our algorithm—the 2-dimensional equivalent of the 3-dimensional algorithm—to 8 iterations. Each iteration consisted of x- and y-translations and a rotation perpendicular to the (planar) imaging plane. Although we assumed that rotations other than those around the axis perpendicular to the imaging plane are negligible, the implications of this assumption were investigated in detail using phantom data.

Data Acquisition and Analysis

Registration accuracy using the described algorithm can be affected by a variety of factors. In addition to the basic evaluation of registration accuracy and precision, we also investigated whether compensating for the photon energy differences between CT and

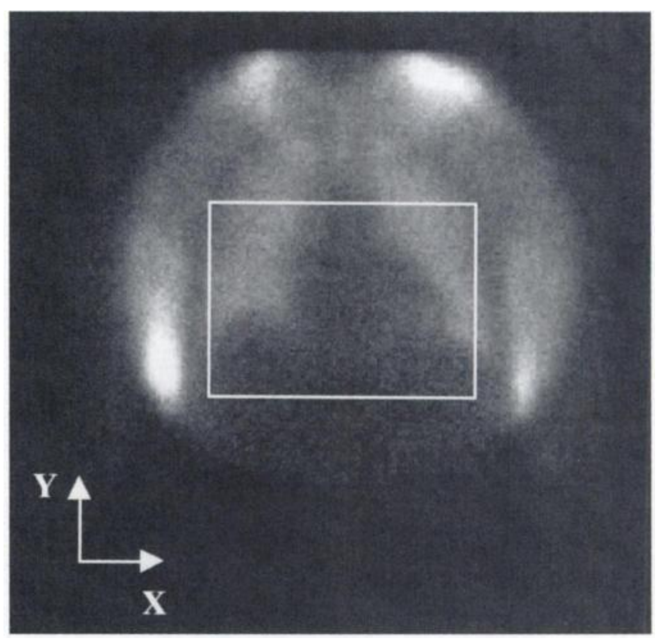

FIGURE 2. Planar transmission scan with typical ROI used for registration calculations. Lower half of lung and upper third of liver were included in ROI; outer body surface contour was excluded.

^{99m}Tc was necessary, and we examined the impact of out-of-plane motions (i.e., yaw and pitch) on the registration results.

Phantom and Patient Data Acquisition

γ-camera transmission and CT data were obtained for an anthropomorphic torso phantom (Data Spectrum Corp., Hillsborough, NC) having a lung insert. Six external fiducial markers were attached to the phantom and were visible in both the planar and the CT datasets. The γ-camera data were acquired using a dual-head Biad camera (Trionix Research Laboratory, Twinsburg, OH) equipped with medium-energy (ME_PAR; Trionix) collimators. Anteroposterior transmission images of the phantom were obtained using a fillable flood source (555 MBq ^{99m}Tc) that was collimated with a high-sensitivity collimator (Siemens Gammasonics, Des Plaines, IL) to reduce transmission-source scatter in the transmission measurement. The source and collimator were placed posterior to the phantom, and the data were acquired for 5 min using the anterior camera. The radiation dose rate from the collimated transmission source was measured to be less than 0.4 mrem (0.004 mSv) per 5-min scan. When needed, downscatter of emission photons into the transmission energy window was compensated for by subtracting an equal-duration downscatter image from the emission-contaminated transmission image. This downscatter image was obtained by acquiring a second "transmission" image with the transmission source removed such that only scattered emission photons were recorded.

The CT data were acquired on a 9800 scanner (General Electric Medical Systems, Milwaukee, WI) with an in-plane pixel size of less than 1 mm and a slice thickness of 10 mm (a typical choice for CT at our institution). The CT data were subsequently resampled to match the pixel size of the planar γ-camera images (2.6 mm/pixel) for the purpose of registration. In addition, data from 8 patients (who were being imaged for clinical reasons) were studied. Each patient underwent CT, performed as described for the phantom, and anteroposterior planar transmission imaging, also performed as described for the phantom (except that no external fiducial markers were used for the human subjects). In 4 of the 8 subjects, transmission scans were also obtained on each of 4 subsequent days, yielding a set of 5 transmission scans per patient. The CT data for both patients and phantom were reprojected along the anteroposterior direction—i.e., pixels along each vertical column of each image were summed—to form the planar rCT images.

CT-to-^{99m}Tc Energy Scaling Measurements

Because the photon energy used in CT (approximately an 80-keV peak) is significantly lower than the 140-keV photon energy of ^{99m}Tc used for the planar transmission scans, the necessity for rescaling the CT data was investigated. An approximate method of rescaling CT data to yield attenuation coefficients for different photon energies has been described (*10,11*). We evaluated the registration of rCT with and without rescaling to determine whether rescaling of the CT data was necessary. We reprojected the CT data from the 8 patient scans into 2-dimensional planar images in 2 ways. First, we reprojected the transaxial CT slices using the original unscaled CT values. Second, we reprojected the same CT data after rescaling them to correspond to the attenuation coefficients for the photon energy of ^{99m}Tc (*10,11*). We also created duplicates of the rescaled CT reprojections, which were then smoothed to a spatial resolution similar to that of the 140-keV transmission images and to which Poisson noise was added at a level similar to 1 of our acquired 140-keV transmission images. These modified CT reprojections served as simulations of nuclear medicine transmission images, which were then deliberately misaligned by known amounts. The registration algorithm was subsequently used to align the simulated images both to the original rCT and also to the rescaled rCT. These data permitted us to determine whether inaccuracies in the rescaling process would adversely affect the alignment results.

Measurements of Precision and Accuracy of Registration

To test the alignment of actual measured 140-keV transmission data with rCT, we examined the CT and planar data obtained for the torso phantom. The 140-keV transmission data were converted to the same units as the rCT by taking the negative natural log of the transmission data. Fiducial markers had been attached to the outside of the phantom. The markers were plastic tubes, having an inner diameter of approximately 2 mm, glued to Styrofoam (Dow Chemical Co., Midland, MI) blocks that were then attached to the outer surface of the phantom. The positions of the markers were chosen so that all possible translations and rotations could be detected and so that the markers fell outside the ROI used by the registration algorithm. The markers were filled with ^{99m}Tc for γ-camera visualization and were also clearly seen on the CT images without a contrast agent. The CT data were reprojected and resampled to have the same pixel size as the 140-keV planar transmission data before registration. To test the dependence of the algorithm on the degree of initial misalignment, a set of measurements was performed in which 2 datasets were first aligned manually using the external fiducial markers. Then, known misalignments were deliberately introduced into the γ-camera transmission data 5 separate times using translations of up to 13 mm and rotations (in the planar imaging plane) of up to 5°. The mean registration errors were determined from the translations and rotations predicted by the registration algorithm as compared with the known movements.

Recognizing that 2-dimensional alignment does not account for all possible 3-dimensional rotations, we also examined the impact of yaw and pitch rotations (Fig. 3). Yaw rotations are those seen as the patient rotates about an axis perpendicular to a transaxial imaging plane, and, similarly, pitch rotations are those seen as the patient rotates in a sagittal view. The torso phantom data were examined after introducing known yaw and pitch rotations to the CT slices before reprojection. The reprojections were then registered to the planar transmission scans (which, of course, had no additional yaw and pitch rotations introduced). The 2-dimensional alignment parameters determined by the registration algorithm were compared with the known parameters as a measure of alignment error.

To further test the accuracy of the registration algorithm we processed the data from the group of 4 patients who each underwent CT once and planar transmission scanning 5 times (on 5 separate days). The actual translations and rotations between scans were not known, so we performed an initial registration of all the transmission scans to the rCT and assumed that the images were then correctly aligned. As shown below, this assumption was not vital. We then deliberately misaligned each planar transmission scan 5 separate times by randomly chosen amounts (up to 13 mm and 5°, as with the phantom), aligned them to the rCT using the algorithm, and compared the alignment parameters predicted by the algorithm with those introduced artificially. Although this test does not determine the precision of alignment, it does test the accuracy

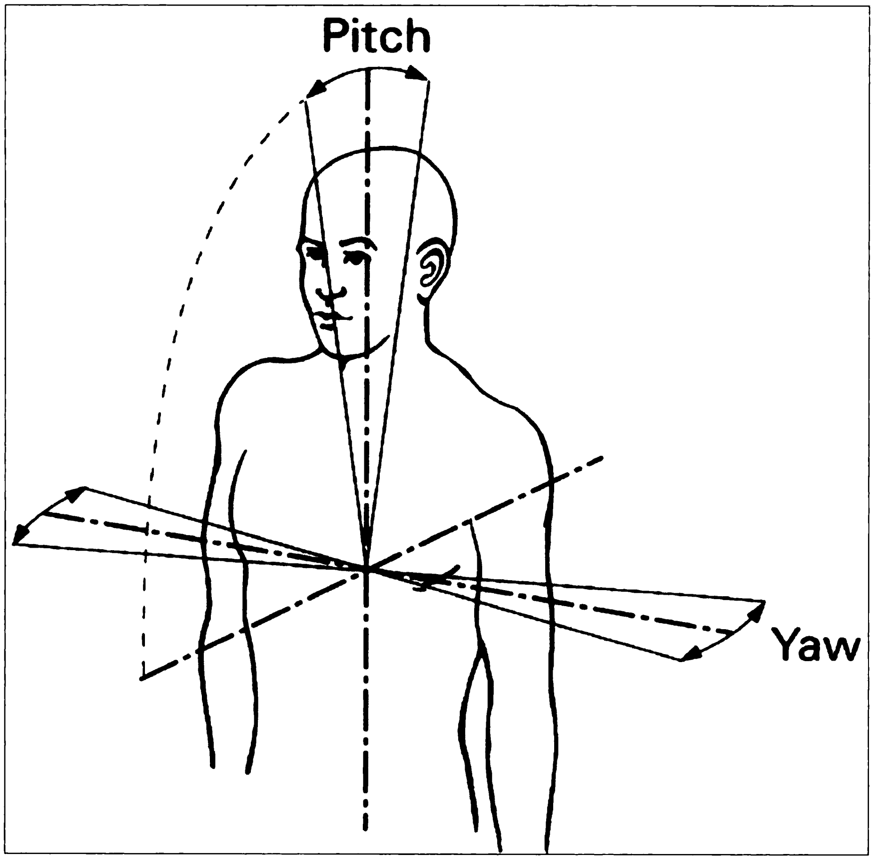

FIGURE 3. Orientations for yaw and pitch rotations.

for a variety of initial misalignments. For each patient, a single ROI was used for alignment of all scans.

Lastly, we investigated the influence of ROI selection on the registration results. We aligned the reprojected CT and planar transmission data from the same 4 patients using 5 ROIs, each of which included the lower half of the lung and the upper third of the liver. Because each patient had 5 transmission scans, this procedure resulted in 25 sets of alignment parameters for each patient. From these 25 sets we computed the SD of the alignment parameters. The mean value of the SDs for all scans was used to estimate the variability caused by ROI selection.

RESULTS

First, we examined the necessity of scaling CT image pixel values to correspond more closely to those of ^{99m}Tc planar transmission images. Reprojections of CT data with and without rescaling were separately registered to 25 realizations of misaligned (up to 13 mm and 5°) simulated planar transmission scans. The registrations obtained were compared with the known misalignments, and the results are shown in Table 1. There were no significant differences between the registration errors obtained from alignments using scaled CT data and those obtained from alignments using unscaled CT data. Both scaled and unscaled data gave alignment errors that were statistically different from 0, but these differences were small (<1 mm). Therefore, we did not scale any CT data for the subsequent analysis.

The algorithm was next used to align rCT of the torso phantom with planar transmission measurements. The true alignments were initially determined using the external fiducial markers. Known translations of up to 13 mm and in-plane rotations of up to 5° were randomly introduced into the planar transmission data 5 separate times. The algorithm was able to determine those misalignments with errors of -0.1 ± 1.0 mm for x-translations, 1.0 ± 1.3 mm for y-translations, and $-0.2 \pm 0.3°$ for rotations (x and y are illustrated in Fig. 2). The effect of out-of-plane rotations was examined by introducing known yaw and pitch rotations into the CT slices before reprojection. As before, known translational and in-plane rotational misalignments were introduced, and the algorithm was applied. The resultant registra-

TABLE 1
Registration Errors for Simulated Planar Transmission Data Aligned to Reprojected CT Data, With and Without Rescaling for Different Photon Energies

Rescaling	x-transmission (mm)	y-transmission (mm)	Rotation (degree)
With	0.2 ± 1.4	0.6 ± 1.8*	0.9 ± 0.8†
Without	0.2 ± 1.6	0.9 ± 2.3*	−0.1 ± 0.3

*Siginficantly different from 0, $P < 0.05$ in *t* test.
†Significantly different from 0, $P < 0.001$.
Value of 0 would indicate perfect alignment by algorithm.

tion errors are shown in Table 2 for several values of yaw and pitch. As can be seen over a range of 10° of yaw and no pitch rotation, the translational alignment errors were, at most, 2 mm. For a pitch angle range of ±2° and no yaw rotation, the translational errors were approximately 4 mm or less.

Next, we registered the CT and transmission data obtained from the subset of 4 patients, each of whom underwent CT once and planar transmission scanning 5 times. Initially, we registered each planar transmission scan to the rCT and considered these datasets to be aligned. We then deliberately misaligned (again by up to 13 mm and 5°) the planar data by known amounts and reapplied the algorithm. Comparing the motions predicted by the second registration with the known introduced motions, we observed registration errors (as measured by the registration differences) of 1.6 ± 0.8 mm for x-translations, 1.3 ± 1.0 mm for y-translations, and 0.5 ± 0.5° for rotations.

Finally, we investigated the influence of ROI selection on the registration error. We processed the data from the same 4 patients by drawing 5 ROIs for each scan. We calculated the SD of the registration differences for each scan and from these determined the mean SD from all 20 transmission scans. The average SDs were 1.0 mm for x-translations, 2.0 mm for y-translations, and 1.2° for rotations.

DISCUSSION

Considerable effort has been expended recently to register SPECT with other modalities. Despite the importance of SPECT imaging, many nuclear medicine images are still acquired in planar format, and planar imaging still plays an important role in noncardiac nuclear medicine. The clinical usefulness of the planar images would greatly improve in many circumstances if they could be aligned to a CT dataset. In some tumor imaging applications, registration could even permit absolute quantitation by furnishing the appropriate attenuation correction factors from the registered CT dataset. In addition, such alignment would be valuable for accurately assessing dosimetry before therapeutic administrations of monoclonal antibodies (*1,2*). At the least, knowledge of the line in the CT slice that corresponds to a particular pixel in the planar emission image provides valuable anatomic correlates to the planar data.

TABLE 2
Registration Errors Resulting from Yaw and Pitch Rotations Introduced into the Torso Phantom CT Slices Before Reprojection

Angle (degree)	x-transmission error (mm)	y-transmission error (mm)	Rotation error (degree)
Yaw			
−5	1.9	1.1	−0.2
−2	0.7	0.1	−0.2
2	−0.4	−0.1	−0.6
5	−1.0	−1.0	−0.9
Pitch			
2	−0.9	−4.1	−0.4
−2	−1.5	−3.0	−1.0

The method of alignment by correlation described here is based on the fact that reprojected and rescaled CT images contain the same information (albeit at higher resolution) as planar nuclear medicine transmission images acquired under the same anatomic conditions. Our evaluation of this method has shown subpixel accuracy (pixel size, 2.6 mm) in a variety of circumstances.

We found that rescaling of the CT pixel values to match the 140-keV ^{99m}Tc photon energy of the planar transmission scan was unnecessary. No significant difference was found between registrations done either with or without rescaling. Both scaled and unscaled data resulted in a small (<1 mm) bias in y-axis alignment, and the scaled data also showed a small (<1°) rotational bias. One can therefore eliminate the cumbersome scaling step. This result was not entirely expected, because the scaling between 140-keV and CT energies is not completely linear. Presumably, this factor reduces overall correlation in the unscaled data but does not alter the spatial location of the maximum correlation.

One might suppose that yaw and pitch differences, introduced by differences in the way patients are positioned on the CT and nuclear medicine tables, reduce the accuracy of the alignment. Such a reduction was, in fact, found to be the case. Pitch rotations (i.e., more or less elevation of the patient's upper body than the patient's lower body) of 2° gave larger errors than did even 5° yaw (i.e., rotations about an inferior–superior line), particularly for y-axis translations. However, even for yaw and pitch differences as great as ±5° and ±2°, respectively, alignment errors of <4 mm were achieved. In practice, keeping the relative yaw and pitch rotations well below these values should be quite easy, thus reducing alignment error further.

The alignment method studied here requires that the user manually place a rectangular ROI around a region including the lower lungs and upper liver, while excluding the lateral external body surfaces. This manual step might be expected to introduce some variability into the method. However, SDs caused by ROI selection were found to be on the order of only 1–2 mm for translations and approximately 1° for rotations. These values resulted in misalignments less than the pixel size used for the registration algorithm. Furthermore, the pixel size in the y-direction for the rCT data was 10 mm rather than 2.6 mm for the planar data. This fact likely contributes to the higher y-translational error observed and suggests that the errors might have been smaller if thinner CT slices had been used.

The intent of this study was to test a method for transmission scan–rCT image alignment under the assumption that the patient's underlying anatomy was similar at the time of the 2 scans. In practice, the CT scan is often acquired during a breath-hold, whereas the transmission scan is acquired during normal respiration. Several methods are available to address this problem. First, a short CT scan without breath holding can be obtained along with the standard diagnostic CT scan. Such scans are routinely performed when CT data are acquired for radiation therapy planning, and CT scans without breath holding have also been used to perform PET–CT fusion (*12–14*). Alternatively, the transmission data can be gated to the respiratory cycle to produce a "breath-hold" transmission scan (*15*). Our purpose was not to investigate which of these methods for respiratory compensation is best but rather to determine whether the concept of rCT–transmission scan alignment can be performed accurately given that acquisitions are performed under similar anatomic configurations. We did, however, make a worst-case estimate of the size of the error that results if one does not account for respiratory motion. To do so, we obtained planar transmission data under different respiratory conditions. A 5-min planar transmission image was obtained during normal patient breathing and was followed by a series of very deep (maximum end-inspiratory) breath-hold planar transmission images that were summed to obtain a 5-min breath-hold image. A second image during normal breathing was then taken after the breath-hold series to check for patient motion. Registration of the summed breath-hold image with the image during normal breathing revealed 2.4 and 11.8 mm of registration difference in the x- and y-directions, respectively, with no rotational motion. As expected, the primary apparent motion was along the inferior–superior line (y-axis), arising from the displacement of the inferior lung borders relative to the rest of the lung. Registration of the 2 images during normal breathing indicated no translational motion and 0.6° of rotational motion. These results indicate that if no respiratory correction is applied, breathing effects can introduce inaccuracies of approximately 12 mm or less in positioning along the inferior–superior line. Consequently, caution is indicated for patients who undergo only deep breath-hold CT and for whom no respiratory compensation has been performed.

Another problem arises if the patient moves between the transmission scan and the emission scan. However, we contend that given appropriate precautions this movement can be minimized to a level that is not clinically significant—a fact supported by the widespread use of pre- or postemission scan transmission scanning in PET. Minimization of movement should be especially true for the relatively short imaging times often used in planar imaging. For example, at our institution we typically acquire 2- to 5-min transmission scans followed immediately by 5- to 10-min emission scans.

Postinjection transmission imaging can be dramatically affected by the presence of emission photon scattering into the transmission photon energy window. Left uncorrected, this downscatter can severely affect the accuracy of lung-based registration. A simple example is the case of ^{99m}Tc transmission imaging performed soon after injection of ^{111}In-labeled antibodies. Because the antibodies remain in the blood pool for many hours, downscatter from ^{111}In in the blood pool into the lung regions adversely affects ^{99m}Tc transmission imaging of the lungs. However, we have found that a simple measurement of emission photons in the transmission energy window provides a robust means of compensating for downscatter. Straight subtraction of an equal-duration downscatter image from the emission-contaminated transmission image gives net registration errors of <1 mm for translations and $<1°$ for rotations.

CONCLUSION

A registration algorithm based on maximizing the correlation coefficient between rCT and planar γ-camera transmission images has been shown to permit alignment of projected CT data with planar transmission images. The accuracy and precision of alignment were good, except when yaw and pitch differences were excessive. When coupled with respiratory corrections, this algorithm should make possible the relation of the anatomic features shown in CT data to the physiologic information shown on planar nuclear medicine images.

REFERENCES

1. Carrasquillo JA, White JD, Paik CH, et al. Similarities and differences in ^{111}In- and ^{90}Y-labeled 1B4M-DTPA antiTac monoclonal antibody distribution. *J Nucl Med.* 1999;40:268–276.
2. Mulligan T, Carrasquillo JA, Chung Y, et al. Phase I study of intravenous Lu-177-labeled CC49 murine monoclonal antibody in patients with advanced adenocarcinoma. *Clin Cancer Res.* 1995;1:1447–1454.
3. Thomas SR, Maxon HR, Kereiakes JG. In vivo quantitation of lesion radioactivity using external counting methods. *Med Phys.* 1976;03:253–255.
4. Bacharach SL, Douglas MA, Carson RE, et al. Three-dimensional registration of cardiac positron emission tomography attenuation scans. *J Nucl Med.* 1993;34:311–321.
5. Barillot C, Lemoine D, Le Briquer L, Lachmann F, Gibaud B. Data fusion in medical imaging: merging multimodal and multipatient images, identification of structures and 3D display aspects. *Eur J Radiol.* 1993;17:22–27.
6. Holman BL, Zimmerman RE, Johnson KA, et al. Computer-assisted superimposition of magnetic resonance and high-resolution technetium-99m-HMPAO and thallium-201 SPECT images of the brain. *J Nucl Med.* 1991;32:1478–1484.
7. Pelizzari CA, Chen GT, Spelbring DR, Weichselbaum RR, Chen CT. Accurate three-dimensional registration of CT, PET, and/or MR images of the brain. *J Comput Assist Tomogr.* 1989;13:20–26.
8. Weber DA, Ivanovic M. Correlative image registration. *Semin Nucl Med.* 1994;24:311–323.
9. Woods RP, Mazziotta JC, Cherry SR. MRI-PET registration with automated algorithm. *J Comput Assist Tomogr.* 1993;17:536–546.
10. Koral KF, Zasadny KR, Kessler ML, et al. CT-SPECT fusion plus conjugate views for determining dosimetry in iodine-131-monoclonal antibody therapy of lymphoma patients. *J Nucl Med.* 1994;35:1714–1720.
11. Nickoloff EL, Perman WH, Esser PD, Bashist B, Alderson PO. Left ventricular volume: physical basis for attenuation corrections in radionuclide determinations. *Radiology.* 1984;152:511–515.
12. Kotz D. Nuclear medicine in the 21st century: integration with other specialties. *J Nucl Med.* 1999;40(7):13N–14N, 21N, 25N–26N.
13. Charron M, Beyer T, Kinahan PE, Meltzer CC, Dachille MA, Townsend DW. Whole-body FDG PET and CT imaging of malignancies using a combined PET CT scanner [abstract]. *J Nucl Med.* 1999;40(suppl):256P.
14. Townsend DW, Beyer T, Kinahan PE, et al. Fusion imaging for whole-body oncology with a combined PET and CT scanner [abstract]. *J Nucl Med.* 1999;40(suppl):148P.
15. Klein GJ, Reutter BW, Ho MH, Reed JH, Huesman RH. Real-time system for respiratory-cardiac gating in positron tomography. *IEEE Trans Nucl Sci.* 1998;45: 2139–2143.