Effects of shortened acquisition time on accuracy and precision of quantitative estimates of organ activity^{a)}

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Purpose: Quantitative estimation of *in vivo* organ uptake is an essential part of treatment planning for targeted radionuclide therapy. This usually involves the use of planar or SPECT scans with acquisition times chosen based more on image quality considerations rather than the minimum needed for precise quantification. In previous simulation studies at clinical count levels (185 MBq ¹¹¹In), the authors observed larger variations in accuracy of organ activity estimates resulting from anatomical and uptake differences than statistical noise. This suggests that it is possible to reduce the acquisition time without substantially increasing the variation in accuracy.

Methods: To test this hypothesis, the authors compared the accuracy and variation in accuracy of organ activity estimates obtained from planar and SPECT scans at various count levels. A simulated phantom population with realistic variations in anatomy and biodistribution was used to model variability in a patient population. Planar and SPECT projections were simulated using previously validated Monte Carlo simulation tools. The authors simulated the projections at count levels approximately corresponding to 1.5–30 min of total acquisition time. The projections were processed using previously described quantitative SPECT (QSPECT) and planar (QPlanar) methods. The QSPECT method was based on the OS-EM algorithm with compensations for attenuation, scatter, and collimator-detector response. The QPlanar method is based on the ML-EM algorithm using the same model-based compensation for all the image degrading effects as the QSPECT method. The volumes of interests (VOIs) were defined based on the true organ configuration in the phantoms. The errors in organ activity estimates from different count levels and processing methods were compared in terms of mean and standard deviation over the simulated phantom population.

Results: There was little degradation in quantitative reliability when the acquisition time was reduced by half for the QSPECT method (the mean error changed by <1%, e.g., 0.9%–0.3% = 0.6% for the spleen). The magnitude of the errors and variations in errors for large organ with high uptake were still acceptable for 1.5 min scans, even though the errors were slightly larger than those for the 30 min scans (i.e., <2% for liver, <3% for heart). The errors over the ranges of scan times studied for the QPlanar method were all within 0.3% for all organs.

Conclusions: These data indicate that, for the purposes of organ activity estimation, acquisition times could be reduced at least by a factor of 2 for the QSPECT and QPlanar methods with little effect on the errors in organ activity estimates. The acquisition time can be further reduced for the QPlanar method, assuming well-registered VOIs are available and the activity distribution in organs can be treated as uniform. Although the differences in accuracy and precision were statistically significant for all the acquisition times shorter than 30 min, the magnitude of the changes in accuracy and precision were small and likely not clinically important. The reduction in SPECT acquisition time justified by this study makes the use of SPECT a more clinically practical alternative to conventional planar scanning for targeted radiotherapy treatment planning.

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I. INTRODUCTIONS

Quantitatively accurate and precise estimation of the organ activity is an essential element of patient-specific dosimetry for targeted radionuclide therapy treatment planning. The conventional method used to estimate the organ activities has been conjugate view planar scans and processing methods

based on the geometric mean (GM) method.¹ Because this method offers the advantage that the results are theoretically independent of the source depth, and modern dual-head SPECT systems provide a convenient means for simultaneous acquisition of the two conjugate views, it is still widely used in research and has been applied to clinical trials

to calculate absorbed dose to tumors and organs.²⁻⁷ However, there are several physical factors that limit the quantitative accuracy of planar imaging-based methods. These factors include attenuation, scatter, collimator-detector response, and partial volume effects. A number of methods have been proposed to improve the quantitative accuracy of the conventional planar processing methods.⁸⁻¹⁷ Even with the addition of these compensations, the accuracy and precision of the conventional planar method is still limited for the following reasons:

- (1) Background subtraction is somewhat subjective and exact compensation for overlap and background activity in the planar projection view is impossible without additional information about the 3D activity distribution.
- (2) The GM method, as conventionally applied, can only partially compensate for attenuation for distributed activity distributions (as opposed to point sources).
- (3) The GM method requires data from a transmission scan for body thickness compensation.
- (4) Energy-window-based scatter compensation methods, often used in conjunction with the GM method, are inexact and can reduce precision due to noise in the scatter window data.
- (5) Compensation for the distant-dependent collimator and detector response (including scatter and septal penetration in the collimator) compensation is difficult and inexact.
- (6) It is difficult to quantify activity in organs with low uptake since they may not be visible in planar images.
- (7) Accurate partial volume compensation is not possible, resulting in poor quantification of small organs and tumors.

In addition to the relatively poor accuracy, we have previously demonstrated that there are substantial variations in accuracy for GM-based planar processing methods due to anatomical and uptake variations.¹⁸

Another alternative for absolute organ activity quantitation is using single photon emission computed tomography (SPECT). By defining volume of interests (VOI) in 3D, it is possible to entirely avoid the problems of organ and background overlap inherent in planar processing methods. It has been demonstrated that, with appropriate SPECT acquisition, calibration, and quantitative reconstruction, SPECT can provide accurate estimates of *in vivo* radioactivity distributions.¹⁹⁻²²

Although quantitative SPECT (QSPECT) may become the method of choice for quantifying *in vivo* activity distribution, it is currently difficult to implement clinically because of the more complex imaging protocol required to cover the dosimetrically important organs, the longer acquisition times conventionally used, and the need to use computationally intensive iterative reconstruction methods. In order to address some limitations of the QSPECT method, we have developed a new quantitative planar (QPlanar) method

which is based on maximum-likelihood estimation of organ activities using 3D organ VOIs and a model-based projector that models image degrading effects including attenuation, scatter, and the full collimator-detector response.²³

We have evaluated the accuracy and precision of organ activity and residence time estimates for the GM-based planar, QSPECT, and QPlanar methods, including the components due to statistical noise²²⁻²⁴ and patient variation,¹⁸ using both simulated and physical phantom experiments. The results showed that for typical clinical count level, the variations in accuracy observed for the phantom population were much greater than the variations due to statistical noise. The variations in accuracy were also substantially larger for conventional GM-based planar quantification methods than for QSPECT or QPlanar methods. Since the variations in accuracy due to patients are a random effect, they are analogous to variations in accuracy due to statistical noise and can be thought of as affecting the precision of the activity estimates. These results suggest that acquisition time, typically chosen based more on visual image quality considerations rather than the minimum needed for reliable quantification, could be reduced without decreasing the overall precision. We thus hypothesize that acquisition times for QPlanar and QSPECT quantification methods can be reduced without substantially reducing the accuracy or increasing the variability of activity estimates.

Ideally, one would validate this hypothesis using patients scanned with clinical imaging systems. However, these studies would be very expensive and difficult to perform as it is not possible to accurately and independently determine the *in vivo* activity in organs. Using animal studies with postmortem measurements of organ activity will also not provide definitive proof because the organ sizes and geometries in animals are quite different than that in humans. Using physical phantom experiments is insufficient because of the lack of variability in terms of anatomy and biodistribution. Thus, in this work, simulated phantom population data were used to evaluate the effects of reducing acquisition time on the accuracy and precision of organ activity estimates. Both the QSPECT and QPlanar methods were evaluated in terms of the accuracy and precision of organ activity estimation.

We did not investigate the effects of reduced acquisition time on the accuracy or precision of conventional GM-based planar quantification for several reasons. First, the accuracy and precision is already inferior to that obtained with QPlanar or QSPECT methods. Second, because conventional planar processing requires definition of organ and background regions of interest (ROIs) directly on the planar images. Thus, the noise in the image will impact the precision not only through the noise, but by variations in the ROIs. For the QPlanar and QSPECT methods investigated we assumed that the VOIs were defined using a registered anatomical image, and thus VOI accuracy was not affected by increased noise in the nuclear medicine images.

II. METHODS

II.A. Phantom population and Monte Carlo simulation

II.A.1. Phantom population

The 3D NCAT phantom²⁵ was used to provide a realistic and flexible model of human anatomy and physiology. Using diagnostic CT and SPECT/CT images from a clinical trial of high dose myeloablative ⁹⁰Y-Zevalin therapy,²⁶ we measured the length, width, and thickness of dosimetrically important organs that had high activity uptake, including the bone marrow, heart, liver, lungs, spleen, and kidneys. These values were used to rescale the sizes of the corresponding organs in the NCAT phantom by shrinking or stretching each organ so it had the same size in these three dimensions. The result was a phantom where the values of these parameters were the same as for the patient, though the exact organ shapes, volumes, and geometric relationships were different. This process was repeated for seven consecutive patients to generate a population of phantom anatomies. The activity concentrations in the major organs (heart, lungs, liver, kidneys, spleen, and bone marrow) of these seven patients were also estimated using SPECT/CT scans at 24 h after injections.²⁶ Detailed organ activity distribution and volume information are given in Ref. 18. Note that the activity distribution in all the organs was uniform with the exception of the lungs. The NCAT phantom includes a model of the airways in the lungs; activity in the airways was set to zero, though they were included as part of the lung VOI.

We used each set of anatomic parameters with each set of organ biodistribution parameters to form a total of 49 sets of parameters. Below we describe how these sets of parameters were used to produce 49 different projection data sets.

II.A.2. Monte Carlo simulation

A modified version of the SimSET/PHG Monte Carlo code²⁷ combined with the angular response function (ARF) method²⁸ was used to simulate the projection data. The ARF method is based on full simulations of the collimator-detector point response function using a modified version of the Monte Carlo N-Particle transport code²⁹ and has been previously validated.^{22,30,31} It allows fast and accurate modeling of the collimator-detector response including the effects of attenuation and scatter in the septa and crystal.

The simulations were performed using parameters appropriate for a GE Discovery VH/Hawkeye SPECT/CT system (General Electric, Milwaukee, WI) with a 2.54 cm thick crystal and a medium energy general purpose collimator (hex holes 5.8 cm long with a flat-to-flat distance of 0.3 cm and a septal thickness of 0.105 cm). Please note that the sensitivity of this camera is higher than cameras with 0.95 cm thick crystals. The simulation modeled a 9.5% energy resolution at 140.5 keV and a 4.5 mm intrinsic spatial resolution. Both ¹¹¹In photopeaks (171 and 245 keV) were separately simulated with the appropriate abundances. Low-noise projection images were generated in 128 transaxial and 170 axial projection bins at 120 views over 360° using a 0.442 cm projection bin size and two 14% wide energy windows centered

at 171 and 245 keV. We simulated a total of $9\text{--}13 \times 10^9$ photon histories for each phantom to generate projections with noise levels much lower than the clinical planar or SPECT projections. Note that because of the ARF method, the required total number of simulated photon for generating low-noise projections is largely reduced (by a factor of hundreds) compared to those required in full Monte Carlo simulations.²⁸ We then used a Poisson distributed pseudorandom noise generator to simulate the desired noise levels in the projection data, as described below.

In order to reduce discretization effects, the activity and material maps were discretized using voxels 1/8 the volume (0.221 cm cubic voxels) of the simulated projection bins (0.442 cm bin size). For each phantom of the seven sets of patient anatomies, simulations were performed separately for each organ. The organs were then scaled and summed, based on the whole-organ activity uptake for a given patient, to form a set of projection data corresponding to the same anatomy but with variations in the biodistribution. This method was used to reduce the number of Monte Carlo simulations required, but to provide a large number of combinations of variations in both biodistribution and anatomies. In this way, using the seven sets of NCAT phantoms and seven sets of organ activity uptakes, we generated 49 sets of projection data corresponding to the realistic anatomies with variations in the biodistribution.

For each set of projection data, the low-noise projections were first scaled to count levels appropriate for clinical planar and SPECT acquisitions at the corresponding injected activities and scan times (30 min scan at 24 h post a 185 MBq ¹¹¹In injections). The average over all slices and all patients of the total counts in the SPECT projection sinograms was 1.02×10^5 (120 views in total). The average of the total counts in each of the anterior planar images was 1.01×10^7 . Note that in the simulation, we assumed the field-of-view of the SPECT was the same as that in the planar scans. In reality, two or three SPECT scans would be needed to cover all the dosimetrically important organs. In the rest of the paper, the SPECT scan times refer to the total acquisition time for one SPECT bed position, and not the time for all the bed positions needed to acquire an equivalent field-of-view as used for the planar projections.

To simulate reduced acquisition times, these projections were then scaled to lower count levels that corresponded to scan times of 15, 9, 3, and 1.5 min. Poisson noise was then simulated using a Poisson random number generator to simulate projections at appropriate noise levels corresponding to the different count levels. Figure 1 shows samples of noisy anterior planar images at five different count levels (left to right) for two sample phantoms (rows 1 and 2) and one view (out of a total of 120) of the SPECT projections at the corresponding count levels. Please note that for the same total scan time, the counts in one view of the SPECT projections was 1/60 of those of corresponding planar images.

II.B. Quantitative processing methods

All the simulated planar images and SPECT projections were processed using the two quantitative processing meth-

FIG. 1. Rows 1 and 2 are anterior projections with count levels corresponding to (left to right) acquisition times of 30, 15, 9, 3, and 1 min for two sample phantoms with different anatomies and biodistributions. The last row shows the anterior SPECT projection (120 in total) for the phantom in row 2 for the same total acquisition times.

ods described below. The percent errors compared to the truth and variation in percent errors over the phantom population (49 phantoms in total) for the various count levels were calculated and compared for these two methods.

II.B.1. QSPECT method

SPECT reconstruction is a promising imaging method for quantitatively estimating the 3D activity distributions *in vivo*. However, SPECT images without compensation for physical image degrading effects, such as attenuation and scatter, are not quantitative in the sense that the reconstructed voxel values are not proportional to the true activities in the voxels. In previous works, we have demonstrated that the QSPECT reconstruction can provide accurate organ activity and residence time estimates using both physical phantom experiment and simulations for both single phantoms and population of NCAT phantoms.^{18,22–24}

The QSPECT method used in those papers and in this work is based on the iterative ordered-subsets expectation-maximization (OS-EM) algorithm³² with attenuation, scatter, and collimator-detector response function compensations. No additional partial volume compensation was performed. For attenuation compensation, the abundance-weighted average of the true attenuation maps for the 171 and 245 keV photons emitted by In-111 was used in the reconstruction. Scatter compensation was performed using a fast implementation of the effective source scatter estimation (ESSE) method.^{33,34} Point sources at various distances from the face of the collimator were simulated to estimate the distance-dependent CDRF and modeled interactions with and penetration through the collimator and detector.³¹ A detailed description and validation of the QSPECT method can be found in Ref. 22. The SPECT projections were reconstructed using the QSPECT method (30 iterations, 24 subsets per iteration) described above. The organ VOIs were defined based on the

true configuration in the phantoms (0.221 cm pixel size). Since the pixel size in the reconstructed images was 0.442 cm, we used nearest-neighbor interpolation to halve the voxel size in each dimension and produce an image with the same dimensions as that of the true organ VOIs. We then computed the total activity in each organ by summing the voxel values in the expanded image that were within the organ VOI.

II.B.2. QPlanar method

Although the QSPECT method can provide accurate and precise organ activity estimates (accuracy better than 5% for most organs as reported in Ref. 22), planar methods are often used clinically because of the complex imaging protocol and longer imaging times conventionally used for SPECT. To address some of the limitations of the conventional planar processing methods while retaining the practical advantages of planar imaging, we have developed the QPlanar processing method.²³ This method uses an iterative maximum-likelihood expectation-maximization (ML-EM) algorithm to estimate the organ activities directly from the anterior and posterior planar images in the following steps:

- (1) Define 3D organ VOIs (we used the true 3D organ configurations in the simulations).
- (2) Model the activity distribution in each VOI (we assumed the uniform distribution inside each VOI in this study).
- (3) Estimate the projection of each separate organ VOI using a projector which includes modeling of image degrading effects.
- (4) Assuming the measured projection is a linear combination of these separate VOI projections, use the ML-EM algorithm to estimate the organ activities needed to scale the VOI projections to provide the best match to the measured data in a maximum-likelihood sense.

A major difference between this method and the 3D SPECT reconstruction is that only a small number of parameters (equal to the number of organs) are estimated, and thus they can be estimated quite effectively from only two projection views. More detailed information about the QPlanar method can be found in Ref. 23.

III. RESULTS

The percent errors in organ activity estimates compared to the true activities and the variation in these percent errors over the phantom population for the five different count levels for QSPECT and QPlanar are shown in Tables I and II, respectively.

For the QSPECT method, the accuracy and precision (as measured by the variation in the errors) of organ activity estimates degraded with decreased acquisition time for all organs, especially for the small organs such as kidneys and marrow. There were small (<1%) differences in organ activity estimates after decreasing the acquisition time from 30 to 15 min. Even though the differences in the activity estimates for shorter scans were statistically significant (paired *t*-test

TABLE I. Percent errors^a and percent variation in errors over the phantom population for various scan times using the QSPECT method.

	<i>Heart</i>	<i>Lungs</i>	<i>Liver</i>	<i>Kidneys</i>	<i>Spleen</i>	<i>Marrow</i>
30 min	$-2.6 \pm 1.6\%$	$-4.4 \pm 6.0\%$	$-0.5 \pm 0.8\%$	$-4.5 \pm 2.7\%$	$0.9 \pm 1.7\%$	$-7.3 \pm 1.8\%$
15 min	$-2.8 \pm 1.8\%$	$-4.8 \pm 6.4\%$	$-0.7 \pm 0.8\%$	$-5.5 \pm 2.9\%$	$0.3 \pm 2.0\%$	$-8.2 \pm 2.5\%$
9 min	$-3.3 \pm 1.9\%$	$-4.9 \pm 6.3\%$	$-1.1 \pm 1.1\%$	$-5.3 \pm 2.8\%$	$0.3 \pm 1.9\%$	$-9.5 \pm 2.9\%$
3 min	$-3.8 \pm 2.7\%$	$-7.1 \pm 7.1\%$	$-1.8 \pm 1.6\%$	$-8.6 \pm 4.2\%$	$-1.9 \pm 3.1\%$	$-12.2 \pm 4.9\%$
1.5 min	$-5.3 \pm 3.2\%$	$-7.8 \pm 6.8\%$	$-2.2 \pm 1.5\%$	$-9.8 \pm 7.4\%$	$-3.9 \pm 4.6\%$	$-15.6 \pm 6.6\%$

^aCalculated as $(\text{estimated activity} - \text{true activity}) / \text{true activity} \times 100\%$. Negative signs indicate underestimation compared to the true activity.

compared to 30 min scans), the magnitude of the errors and variations were still acceptable for most organs, especially for large organs such as the liver.

At first glance, it might seem surprising that the accuracy of QSPECT was so robust with respect to reduced acquisition time. However, it should be remembered that for an unbiased estimator, accuracy is independent of noise level. Further, if an unbiased estimator exists, the maximum-likelihood estimator is unbiased. However, due to the ill-posed nature of the voxel reconstruction problem in SPECT, maximum-likelihood reconstruction is not unbiased on the voxel level. In addition, OS-EM is not guaranteed to find the maximum-likelihood solution. Nevertheless, organ activities for large organs such as the liver are very likely to be well estimated, and thus the insensitivity to noise, especially for noise levels where OS-EM approaches the maximum-likelihood solution, is theoretically explicable.

The relatively small increase in variability of activity estimates might also seem puzzling given the factor of 20 decreases in acquisition time. However, recall that the variability reported in Table I includes variability due to anatomical and uptake variations. The total standard deviation can be approximated as the quadrature sum of the standard deviations due to Poisson counting statistics and phantom variations. Previous data indicates that for the 30 min acquisitions the component of the standard deviation due to anatomical variability is larger than that due to noise. For example, the standard deviation for the liver over 50 noise realizations and a single phantom was 0.4%.²³ From Table I, the total standard deviation for the liver was 0.8%. Thus the component of standard deviation due to phantom variations is $\sqrt{0.8\%^2 - 0.4\%^2} = 0.69\%$. The noise standard deviation should

increase in inverse proportion to the change in acquisition time. So, for the 1.5 min acquisition time we would expect that the noise component of the total standard deviation would be $\sqrt{20} \times 0.4\% = 1.8\%$. Thus, the total standard deviation for this time point would be $\sqrt{0.69\%^2 + 1.8\%^2} = 1.9\%$. Given the simplicity of this calculation, this compares well with the 1.5% total standard deviation shown in Table I. In summary, the reason that the loss of precision is relatively small is that the times in this table represent a transition from a regime where the variability is dominated by anatomic variability to one where it is dominated by noise.

For the QPlanar method, the changes in accuracy (mean percent error) in activity estimates for all organs and all the scan times investigated were small ($<0.3\%$). Compared to results from the QSPECT method, QPlanar was less sensitive to the count level. For example, even for the small organs there was a relatively modest loss of precision (increase in percent variation in errors) for the 1.5 min scan time. To investigate the point at which noise in the projection data starts to degrade precision, we continued to reduce acquisition times of 0.6 s. The results are shown in Table III. For the shortest acquisition times there was a loss of precision even for the largest organs.

IV. DISCUSSION

In previous works, we have evaluated the contribution to accuracy and precision of different quantitative activity and residence time estimation methods due to statistical noise²²⁻²⁴ and patient variations.¹⁸ The results showed that the variations in errors for the quantitative imaging methods calculated over a phantom population were much larger than

TABLE II. Percent errors^a and percent variation in errors over the phantom population for various scan times using the QPlanar method.

	<i>Heart</i>	<i>Lungs</i>	<i>Liver</i>	<i>Kidneys</i>	<i>Spleen</i>	<i>Marrow</i>
30 min	$-2.8 \pm 1.8\%$	$12.9 \pm 2.1\%$	$0.11 \pm 0.61\%$	$-5.2 \pm 3.3\%$	$3.1 \pm 1.3\%$	$-5.7 \pm 6.1\%$
15 min	$-2.8 \pm 1.8\%$	$12.9 \pm 2.0\%$	$0.10 \pm 0.58\%$	$-5.1 \pm 3.6\%$	$3.1 \pm 1.2\%$	$-5.8 \pm 5.8\%$
9 min	$-2.8 \pm 1.8\%$	$12.8 \pm 2.1\%$	$0.16 \pm 0.59\%$	$-5.2 \pm 3.5\%$	$3.1 \pm 1.3\%$	$-5.8 \pm 6.1\%$
3 min	$-2.8 \pm 1.8\%$	$12.8 \pm 2.1\%$	$0.11 \pm 0.64\%$	$-4.9 \pm 3.7\%$	$2.9 \pm 1.3\%$	$-5.8 \pm 6.0\%$
1.5 min	$-3.0 \pm 1.8\%$	$13.1 \pm 2.0\%$	$0.13 \pm 0.72\%$	$-5.4 \pm 4.6\%$	$3.2 \pm 1.6\%$	$-6.0 \pm 6.6\%$

^aCalculated as $(\text{estimated activity} - \text{true activity}) / \text{true activity} \times 100\%$. Negative signs indicate underestimation compared to the true activity.

TABLE III. Percent errors^a and percent variation in errors over the phantom population for very short scan times using the QPlanar method.

	<i>Heart</i>	<i>Lungs</i>	<i>Liver</i>	<i>Kidneys</i>	<i>Spleen</i>	<i>Marrow</i>
36 s	$-1.8 \pm 1.9\%$	$10.9 \pm 2.8\%$	$0.8 \pm 1.1\%$	$11.8 \pm 6.6\%$	$0.8 \pm 2.5\%$	$11.2 \pm 8.3\%$
18 s	$1.3 \pm 2.3\%$	$10.9 \pm 2.9\%$	$0.9 \pm 1.2\%$	$12.0 \pm 6.8\%$	$0.9 \pm 2.2\%$	$11.9 \pm 8.6\%$
9 s	$1.4 \pm 2.9\%$	$11.3 \pm 3.4\%$	$1.0 \pm 1.6\%$	$12.7 \pm 9.5\%$	$1.6 \pm 3.3\%$	$12.0 \pm 12.0\%$
1.8 s	$2.4 \pm 5.3\%$	$11.0 \pm 7.4\%$	$0.7 \pm 2.3\%$	$11.5 \pm 18.3\%$	$1.5 \pm 7.0\%$	$11.1 \pm 19.3\%$
0.6 s	$1.4 \pm 11.9\%$	$9.7 \pm 11.9\%$	$0.5 \pm 4.7\%$	$16.1 \pm 31.6\%$	$0.0 \pm 10.1\%$	$16.7 \pm 33.2\%$

^aCalculated as (estimated activity – true activity)/true activity $\times 100\%$. Negative signs indicate underestimation compared to the true activity.

the variations due to statistical noise. This suggests that acquisition time could be reduced without increasing the variation in accuracy over a patient population. In this work, simulated phantom population data were used to evaluate the effect of reducing acquisition time on the accuracy and precision of organ activity estimates. Both the QSPECT and QPlanar methods were evaluated. Although the size of the phantom population (seven anatomies and seven biodistributions permuted to form a set of 49 phantoms) was limited, the results demonstrated the possibility of reducing the acquisition time without increasing the bias and variation in the organ activity estimates.

The reduced acquisition times had little effect on the precisions of the QPlanar data, in contrast to the QSPECT method. This is likely because the number of parameters estimated by the QPlanar method (equal to the number of organs) was much smaller than that in QSPECT method (i.e., number of voxels). The QPlanar and QSPECT methods investigated here use the exact same models of the imaging system, and thus the accuracy of compensation should be similar. One major difference is that the QPlanar method directly estimates the organ activities, which are the parameters of interest in this work. By contrast, when using QSPECT for organ activity quantification, the voxel values are first estimated and then summed to provide the activity estimates. The voxel values are thus nuisance parameters and, in general, the presence of nuisance parameters can increase the variance in an estimation procedure.³⁵ Since there are a large number of nuisance parameters ($\sim 2.7 \times 10^6$), it is thus likely in theory, and we have observed in fact, that organ activity estimation using QSPECT would be more sensitive to noise than using QPlanar.

Another significant difference between QPlanar and QSPECT activity estimation is that QPlanar uses a much less general model of the organ activity distribution, i.e., that the activity distribution is uniform within each organ. Since the activity distribution in the phantom used in this work was uniform for all organs except the lungs, it is not surprising that the QPlanar method was able to achieve high levels of accuracy. It is likely that the results for the QPlanar method would not be as good if the organ activity distributions were nonuniform. One clue to this might be the level of error in the lungs, which was higher than in other organs. However, there are other possible causes for relatively poor accuracy in

the lungs including the close proximity of high activity organs such as the lungs, liver, and blood vessels.

The optimal acquisition time depends critically on the task to be performed with the image. In the above experiment, we assumed that the organ boundaries were known so that the only task performed with the SPECT or planar images was quantification of organ activities. Reducing the acquisition time will result in increased image noise in the reconstructed SPECT images. Thus, if these images are used for visual interpretation, including definition of organ VOIs, the above results may overstate the potential for reduced acquisition time. While studying this effect is beyond the scope of this paper, in the following we address this issue qualitatively by presenting a series of reconstructed images obtained with the data from the different scan times.

It should be kept in mind that the optimal images for quantitative and visual interpretation tasks are, in general, different. We have previously shown that the optimal images for organ activity quantification are obtained for large numbers of iterations (>30).²² For visualization, where quantitative accuracy is not important, a smaller number of iterations and the use of postreconstruction filtering are likely to produce improved image quality, although the accuracy of organ and tumor edges may be compromised. Figure 2 shows samples of reconstructed coronal slices using the various scan times investigated above. Rows 1 and 2 are without postfiltering after the fifth and 30th iteration, respectively. Row 3 is the result in row 2 postfiltered using a 3D Butterworth filter (order 5 and a cutoff frequency of 23 cm^{-1}). Note that the cutoff frequencies used in row 3 were for demonstration only and were not optimized. These images indicate that, while there is clearly degradation in visual image quality with decreasing noise levels, with appropriate regularization it may be possible to use reduced acquisition times for both quantitative and visual interpretation tasks.

Note that in Fig. 2, all the images are displayed on individual gray scales, with the highest voxel value in each image mapped to white. The voxel values with the same gray level are much higher in the second row than those in the first row due to the very intense noise spikes in the images. However, the total organ activities (proportional to the sum of voxel value in organ VOIs) in row 1 are very similar to

FIG. 2. Samples of the reconstructed coronal slices at five different count levels (30–1.5 min from left to right) at the fifth iteration (row 1), the 30th iteration without filtering (row 2), and with a 3D Butterworth (order 5 and a cutoff of 23 cm^{-1}) postfiltering (row 3).

those in row 2. The high noise spikes that become more apparent with iteration are typical of maximum-likelihood type image reconstruction methods.

In many clinical protocols for targeted radiation therapy treatment planning, traditional planar methods have been used even though there is increasing evidence of the superiority of SPECT. Part of the reason for this may be tradition, but part of it may also be because of the perception that SPECT requires long imaging and processing times. Because of the long acquisition time, complex imaging protocols (requiring imaging at multiple bed positions), and longer reconstruction times required to obtain accurate estimates, QSPECT methods have been considered clinically impractical, even though they have been demonstrated to have advantages in providing accurate and precise organ activity and residence time estimation.^{2,3,18,24} The results from this study have shown that it is possible to reduce the total SPECT acquisition time to the same time clinically used for whole body scans. With advances in the computer hardware and the implementation of spiral SPECT capability, QSPECT methods may become more clinically acceptable.

Although it may also be possible to reduce the imaging time for GM-based planar quantification methods, the definition of organ and background ROIs would be difficult on the noisy whole body images, especially for organs with low uptakes. Variation in the definitions of ROIs would have a significant impact on the precision and it is hard to assess these errors. Detailed evaluation of the effects of shortened scan times on GM-based methods is beyond the scope of this paper.

It is important to address some of the limitations of this study and their implications in terms of the clinical applicability of the results. First, the scatter compensation method used in this work was based on modeling the scatter in the projector using ESSE.^{33,34} This method is not generally available clinically. An alternative would be to use energy-

window-based scatter estimation methods. As the acquisition time is decreased, the noise in these scatter estimates would increase, though this could be partially mitigated by low-pass filtering the scatter estimates. This additional noise would, to some degree, degrade the precision of the activity estimates. However, given that the portion of the variation in errors due to Poisson noise is small, it is expected that the effects of noise in the scatter estimates would be relatively small for a decrease in the acquisition time by a factor of 2.

A second limitation of this work is the use of a 25.4 mm thick camera. The efficiency of a camera with a 9.5 mm thick crystal would be about 70% of the one used in this study. This could be taken into account by dividing the acquisition times in the data presented by 0.7. The data would then give an idea of the tradeoff in acquisition time and accuracy that could be obtained with such a camera.

A third limitation is that the calculations were performed for only a single time point representing the activity 24 h postinjection. In typical pretherapy dosimetry studies data are acquired for a period of 7 or more days. There would thus be significantly more noise in the data at the later time points. One approximate way to take this into account would be to divide the acquisition times by the physical decay factor for other time points, though this does not take into account changes in activity distribution due to biological factors (including excretion) that occur over time. As an example, consider an acquisition 144 h postinjection. For this time point, physical decay of ^{111}In would reduce the activity to approximately 29% of the activity at 24 h. Thus, the count level of a 30 min scan at 144 h would approximately equal to that of a 9 min scan at 24 h. The data in Table I suggest that the keeping the 30 min time scan the same would result in relatively modest changes in accuracy and precision, so a 30 min scan might be appropriate at 144 h while the scan time at 24 h could be reduced to 9 or 15 min. However, this does not take into account the effects on the ultimate dose estimates.

A final question is whether, in light of its substantially greater robustness to noise, QPlanar quantification might be a better choice than QSPECT. Despite its appeal in terms of noise robustness, there are a number of practical issues related to the QPlanar method which have not been addressed in this study. For example, QPlanar processing requires 3D regions of interest which are registered to the planar study. Often, these would be obtained from a CT or SPECT study obtained at another time point, and would thus need to be registered. Having substantially noisier projection data might make the registration more difficult and thus increase the sensitivity of the QPlanar method to noise. Another limitation of the QPlanar method is the requirement for a model of the organ activity distribution. In this work, we have assumed that the activity distribution was uniform. In clinical data this assumption might not be sufficiently accurate and might require dividing organs or regions into subregions. This would increase the number of parameters estimated and likely make the QPlanar estimation problem more noise sensitive. Thus, both of these effects would likely make the QPlanar method more sensitive to increased noise in the pro-

jection data. However, the results of this study indicate that the QPlanar method has a number of desirable characteristics, including robustness to noise, that indicate that further development and investigation of it is worthwhile.

V. CONCLUSIONS

In this work, we used a simulated phantom population to investigate the effects of reduced acquisition time on the accuracy and precision of organ activity estimates. The results show that the acquisition time can be reduced at least by a factor of 2 for the QSPECT and QPlanar processing methods with only small changes ($<1\%$) in precision and accuracy of organ activity estimates. The acquisition time can be further reduced for the QPlanar method, assuming well-registered VOIs are available and the activity distribution in organs of interest can be treated as uniform. Although the differences in accuracy and precision were statistically significant for these shorter acquisition times, the magnitude of the changes in accuracy and precision were small and likely not clinically important. The reduction in SPECT acquisition time justified by this study makes the use of SPECT a more clinically practical alternative to planar scanning for targeted radiotherapy treatment planning.

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