Optimization of Energy-Window Settings for Scatter Correction in Quantitative ^{111}In Imaging: Comparison of Measurements and Monte Carlo Simulations

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ABSTRACT

Activity quantification in nuclear medicine imaging is highly desirable, particularly for dosimetry and biodistribution studies of radiopharmaceuticals. Quantitative ^{111}In imaging is increasingly important with the current interest in therapy using ^{90}Y radiolabeled antibodies. One of the major problems in quantification is scatter in the images, which leads to degradation of image quality. The aim of this study was to optimize the energy-window settings for quantitative ^{111}In imaging with a camera that enabled acquisition in three energy windows. Experimental measurements and Monte Carlo simulations, using the SIMIND code, were conducted to investigate parameters such as sensitivity, image contrast, and image resolution. Estimated scatter-to-total ratios and distributions, as obtained by the different window settings, were compared with corresponding simulations. Results showed positive agreement between experimental measurements and results from simulations, both quantitatively and qualitatively. We conclude that of the investigated methods, the optimal energy-window setting was two windows centered at 171 and 245 keV, together with a broad scatter window located between the photopeaks.

Key words: Monte Carlo, dosimetry, quantification, scatter, simulation, ^{111}In imaging

INTRODUCTION

Quantitative ^{111}In imaging is becoming increasingly important, with the current interest in radionuclide therapies using ^{90}Y -labeled radiopharmaceuticals, and it is widely accepted that accurate quantitative imaging is necessary for the

dosimetry for such a targeted radionuclide therapy. A major source of error in quantification is scattered radiation, leading to degradation in image quality and quantitative information.^{1,2} To compensate for this, subtraction of the estimated scatter contribution from the measured projections is a widely used correction method.^{3,4} The scatter distribution for a given radionuclide is, however, dependent on its emitted photon energies. To ensure good image quality, the energy-window setting used should, therefore, be optimized separately for each radioisotope, taking into account the detected energy spectrum. Because scattered photons will be detected in the

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main energy window, the position of additional energy windows for accurate scatter estimation is essential.

The aim of this study was to evaluate different energy-window settings and scatter correction methods for ^{111}In imaging, using a combination of physical phantom experiments and Monte Carlo simulations. The simulations were used to produce reference images comprised of purely primary photons and to estimate parameters that cannot be measured experimentally, such as the fraction of photons that are scattered in a phantom and their contribution to the image.¹ The accuracy of various energy-window scatter correction methods were investigated by comparing the scatter-compensated phantom images with the reference images.

MATERIALS AND METHODS

Phantoms

Three sets of phantoms were used in the experiments and were also created digitally for use in the Monte Carlo simulation. First, image contrast, profile accuracy, and visual examination of the scatter distribution was assessed using a heterogeneous phantom (Fig. 1). This phantom consisted of a large Perspex (PMMA) cylinder with four cylindrical inserts. The inserts consisted of long cylinders of Teflon (PTFE) and water- and air-filled Perspex, representing the spine, liver,

Figure 1. Heterogeneous phantom used for contrast measurement, profile accuracy, and visual evaluation of scatter distribution.

Table 1. Volume and Activity Concentrations of ^{111}In in the Heterogeneous Phantom

	Volume (cm^3)	Activity concentration (MBq/mL)
Background	4554	8.0E-3
Liver	281	5.1E-2
Tumor	50	9.3E-1

and lung, respectively. A smaller cylindrical insert of Perspex represented a tumor. The phantom was filled with activity to mimic the biodistribution in clinical data (see Table 1).⁵ Second, for the sensitivity and scatter-to-total (S/T) estimations, a 3400 cm^3 water-filled cylindrical Perspex phantom with a 6 cm^3 cylindrical insert was used. The insert was filled with 10 MBq of ^{111}In and placed in the center of the larger cylinder. Third, for image resolution measurements, a 1-mm -diameter line source, placed between two 6 cm thick Perspex blocks, was used.

Experimental Acquisitions

Planar images on a dual-head Philips FORTE scintillation camera (Royal Philips Electronics, The Netherlands), with the heads 180 degrees apart, were acquired in list mode so that spectrum files could be produced. The list mode data from each detector was processed separately, using the Royal Marsden Dosimetry Package (RMDP).⁶ Approximately 2 million counts were acquired for the homogeneous phantom and 7 million for the heterogeneous phantom, and transformed into 128×128 matrices with a pixel size of 4.7 mm . Because this was an investigation of scatter correction techniques, it was not necessary to perform an attenuation correction.

Energy-Window Settings

Four combinations of window settings were evaluated: (1) a 20% energy window over the 171 keV photopeak together with two adjacent 6% scatter windows (TEW_171); (2) a 20% energy window over the 245 keV photopeak together with two adjacent 6% scatter windows (TEW_245). Window settings TEW_171 and TEW_245 are conventional triple-energy window scatter corrections, where the number of scattered photons in the photopeak is determined using the method described by Ogawa et al. (Fig.

2A–B).⁷ (3) 20% energy windows over each photopeak together with a 6% scatter window at 149 keV (LEW). The assumption was made that there were no scattered events above the upper limit of the 245 keV window. A line was drawn between this point and the mean counts per energy channel in the scatter window at the lower limit of the 171 keV window (Fig. 2C). Consideration of the scatter contribution from these windows leads to the following equation for the scatter image, $IMA_{scatter}^{LEW}$, to be subtracted from the summed image from the photopeaks:

$$IMA_{scatter}^{LEW} = \left(IMA_{149} \cdot \frac{w_{171}}{w_{149}} \cdot 0.85 \right) + \left(IMA_{149} \cdot \frac{w_{245}}{w_{149}} \cdot 0.22 \right)$$

where IMA_{149} is the image from the scatter window and w_{171} , w_{245} , and w_{149} are the width of the main windows and the scatter window, respectively; and (4) 20% energy windows over each photopeak with a 10% scatter window at 209 keV (EXT). Again, the assumption was made that there were no scattered events above the upper limit of the 245 keV window. A line was drawn between this point and the mean counts per energy channel in the scatter window at the lower limit of the 245 keV window. This line was extrapolated back to the 171 keV window (Fig. 2D). Consideration of the scatter contribution from these windows leads to the following equation for the scatter image, $IMA_{scatter}^{EXT}$, to be subtracted from the summed image from the photopeaks:

$$IMA_{scatter}^{EXT} = \left(IMA_{209} \cdot \frac{w_{171}}{w_{209}} \cdot 2.02 \right) + \left(IMA_{209} \cdot \frac{w_{245}}{w_{209}} \cdot 0.50 \right)$$

where IMA_{209} is the image from the scatter window and w_{209} is the width of the scatter window.

Monte Carlo Simulations

The Monte Carlo simulation code SIMIND was used to simulate projection data under the same conditions as the experiments.⁸ SIMIND was set up to describe the Philips FORTE scintillation camera equipped with an ADAC medium-energy general-purpose collimator (Royal Philips Electronics). SIMIND is based on the use of uniformly distributed random numbers for modeling the different photon interactions. It simulates the passage of photons through a phantom and collimator toward the detector crystal, and calculates the resulting projection images. The detection of photons is affected by attenuation, scatter, septal penetration in the collimator, characteristic X-ray production, and camera characteristics such as intrinsic energy and spatial resolution and the thickness of the scintillation crystal.⁹

For all window settings, simulated “total” and “primary” images were created. The “total” images consist of both primary and scattered photons (just as for the measurements). The “primary” image contains only the primary photons and, therefore, represents the case of a perfectly scatter-corrected image. These primary images will be referred to as reference images for the remainder of this paper. By subtracting the “primary” images from the “total” images, “scatter” images were obtained.

Evaluation

Image quality was assessed by comparisons of sensitivity, image contrast, image resolution, scatter-to-total (S/T) ratios, profile accuracy, and visual evaluation of the scatter distributions.

Sensitivity

The sensitivity was calculated as $Sens = (C_{tot}/t_{acq})/A_t$, where C_{tot} is the summed counts in the whole image, t_{acq} is the acquisition time, and

A_t is the activity within the insert at the time of acquisition.

Image contrast

To measure the contrast, two square regions of interest (ROIs) were placed over the inserts mimicking the tumor and liver. The same ROI was used to analyze all images. The contrast of the “tumor” to the “liver” was calculated from $c = (m_t - m_l)/(m_t + m_l)$, where m_t is the mean pixel value in the tumor ROI and m_l is the mean pixel value in the liver ROI.

Image resolution

To investigate the impact of scatter correction on the spatial resolution, line profiles were obtained and the full width at half maximum (FWHM) was calculated from each of the scatter-compensated images and compared to the matching profiles from the reference images.

Visual evaluation and profile accuracy

Visual evaluation of the various scatter correction methods was performed by comparing the subtracted scatter images to the reference scatter images. In order to estimate the resolution of the structures in the heterogeneous phantom in the scatter-corrected images, line profiles were obtained and compared to the profiles of the reference images (Fig. 3). The profiles were normalized to the maximum pixel value.

Scatter-to-total ratio

The S/T values were calculated as the number of scattered events (S) (subtracted in the scatter correction) divided by the total number of events (T) in the image prior to scatter correction. These ratios were compared to the reference S/T values given by SIMIND.

Table 2. Sensitivity for the Reference Images and the Scatter-Corrected Experimental and Simulation Images

Window setting	cps/MBq Reference	Experimental	Simulation
TEW_171	20.0 ± 7.8E-07	19.2 ± 0.8 (−4%)	19.6 ± 7.9E-07 (−2%)
TEW_245	15.8 ± 8.8E-07	15.5 ± 0.7 (−2%)	16.3 ± 8.8E-07 (+3%)
LEW	35.8 ± 5.6E-07	32.2 ± 1.5 (−10%)	33.1 ± 6.0E-07 (−7%)
EXT	35.8 ± 5.6E-07	35.4 ± 1.5 (−1%)	37.8 ± 5.8E-07 (+6%)

Note. The values in parentheses are the percentage of deviation from the sensitivity in the reference image. Negative signs indicate too low a value, compared to the reference value.

Table 3. References Scatter-to-Total (S/T) Values Obtained from Simulations and Estimated S/T Values for the Different Scatter Correction Methods

Window setting	Reference S/T value	Estimated S/T values
TEW_171	0.50	0.52 (+ 5%)
TEW_245	0.28	0.33 (+ 22%)
LEW	0.42	0.49 (+ 18%)
EXT	0.42	0.44 (+ 6%)

Note. The values in parentheses are the percentage of deviation from the reference S/T value. Positive signs indicate an overestimation of the scatter.

RESULTS

For all evaluated parameters, the experimental and simulation data were found to be in good agreement in most cases. On average, the image contrast values were within 1%, the sensitivity values within 2%, the S/T values within 9%, and the image resolution values within 14%. Generally, good agreement was seen between the spectra obtained from experimental list mode acquisition and simulation (Fig. 4). Although the simulation underestimates characteristic X-ray signal, excellent agreement was observed in the energy range encompassing the two photopeaks that is important for image formation.

It was found that scatter correction improves

both image contrast and image resolution, with values close to the reference. However, there was no significant difference between the results from the four different scatter correction methods. The contrast values were approximately 0.68–0.73 for the images from the photopeaks individually and added together before scatter correction, and slightly increased to 0.76–0.79 after scatter correction. The FWHM values evaluated from the line-source measurements were found to be between 10.1 and 10.5 mm before scatter correction and between 8.7 and 9.3 mm after scatter correction.

The results for sensitivity show that using both photopeaks approximately doubles the sensitivity when compared to using one photopeak alone

Figure 5. The top row shows the reference scatter images in the photopeak energy windows. The bottom row shows the scatter images subtracted with the four different window settings. (A) Reference scatter image for 171 keV. (B) Reference scatter image for 245 keV. (C) Reference scatter image for 171 and 245 keV combined. (D) Reference scatter image for 171 and 245 keV combined. (E) Subtracted scatter image with TEW_171. (F) Subtracted scatter image with TEW_245. (G) Subtracted scatter image with LEW. (H) Subtracted scatter image with EXT.

Figure 6. A posterior abdomen scan of a patient injected with ^{111}In -Zevalin (A) before scatter correction and (B) after scatter correction.

(see Table 2). From a clinical point of view, high sensitivity may reduce the acquisition time or increase the signal-to-noise ratio for a given acquisition time.

In a comparison of the reference S/T ratios provided by SIMIND to the ratios, as estimated by the four different scatter correction methods to the experimental images, it was apparent that the window settings TEW_245 and LEW overestimated the scatter to a greater extent than TEW_171 and EXT (see Table 3). This was also found when visually comparing the subtracted scatter images to the reference scatter images (Fig. 5).

Profiles taken over the heterogeneous phantom indicated that image detail was blurred before scatter correction (Fig. 3). It was also apparent that LEW removed excessive information in the surroundings of the tumor. EXT exhibited a profile that was very close to the reference profile.

DISCUSSION AND CONCLUSION

Using both experimental and Monte Carlo simulated data, a comparison of four different window settings, potentially useful for quantitative ^{111}In imaging, was presented in this paper. The improvement in image contrast and resolution was similar for all four scatter correction techniques and were, therefore, not conclusive parameters. For clinical purposes, a high sensitivity is desirable, which is obtained when using both photopeaks simultaneously. Of the two scatter correction methods that sum data from both photopeaks,

the extrapolated scatter correction technique, EXT, was shown to be superior in a clinical setting, where high sensitivity is of particular importance. This was concluded from evaluating the S/T ratio and the distribution of scatter by the profile analysis. Two 20% photopeak windows, centred at 171 and 245 keV, together with a 10% scatter window at 209 keV, is, therefore, recommended as the optimal energy window setting for quantitative ^{111}In imaging. This window setting has been used in the clinic at The Royal Marsden Hospital (Sutton, United Kingdom) and has been shown to improve the quality of patient images (Fig. 6).

Comparing experimental data with Monte Carlo simulations showed a good agreement. These results validate the use of this program for further investigation, including the potential benefit of adding more energy windows and applications to other radionuclides.

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