Published in final edited form as:

Phys Med Biol. 2011 March 7; 56(5): 1397–1414. doi:10.1088/0031-9155/56/5/012.

Iterative deconvolution of simultaneous ^{99m}Tc and ^{201}Tl projection data measured on a CdZnTe based cardiac SPECT scanner

Krzysztof Kacperski^{1,2}, Kjell Erlandsson¹, Simona Ben-Haim¹, and Brian F. Hutton¹

¹Institute of Nuclear Medicine, University College London, London, UK ²The Maria Skłodowska - Curie Memorial Cancer Center and Institute of Oncology, Warsaw, Poland

Abstract

We present a method of correcting self-scatter and crosstalk effects in simultaneous technetium-99m/thallium-201 stress/rest myocardial perfusion SPECT scans. The method, which is in essence a hybrid between the TEW method and scatter modelling, is based on a model of spatial and spectral distribution of projection counts in several selected energy windows. The parameters of the model are determined from measurements of thin rod sources in air when no in-object scatter or attenuation effects are present. The model equations are solved using the iterative Maximum Likelihood Expectation Maximisation (ML-EM) algorithm in the projection space to find estimates of the primary photopeak counts of both radionuclides. The method has been developed particularly for a novel dedicated cardiac camera based on CdZnTe pixellated detectors, although it can also be adapted to a conventional scintillator camera. The method has been validated in anthropomorphic phantom experiments. Significant improvement in defect contrast has been observed with only moderate increase in image noise. The application of the method to patient data is illustrated.

1. Introduction

The possibility of simultaneous imaging of multiple tracers labeled with different radionuclides is one of the strengths of Single Photon Emission Computerised Tomography (SPECT) which still awaits to reveal its full potential. Radionuclides can be distinguished in the projection data by setting up multiple energy windows which separate photons belonging to different photopeaks. This is not possible for Positron Emission Tomography (PET) where annihilation photons always have the energy of 511 keV, whatever positron emitting radionuclide is used. However, since standard gamma cameras have rather modest energy resolution of the order of 10%, a significant cross-talk of higher energy photopeaks to lower energy windows is typically observed, mainly due to in-patient down-scatter.

Simultaneous dual radionuclide SPECT imaging has several important advantages, particularly when two distinct functional images of the same patient need to be directly compared. One of the most important examples is myocardial perfusion imaging where perfusion in the myocardium is assessed at rest and stress conditions (Berman et al. 1994). By using two tracers labeled with different radionuclides, e.g. ^{99m}Tc and ^{201}Tl , the two images can be acquired simultaneously and identified due to different energies of the gamma photons emitted by the radionuclides. In this way the total time of the procedure can be substantially reduced, improving patient comfort and increasing throughput. Moreover, the two images are perfectly registered with any postural effects or motion artefacts affecting

both in the same way, which avoids frequent difficulties in interpreting images obtained in two separate scans. The main problem in a simultaneous acquisition procedure is the crosstalk between the two radionuclides, primarily due to down-scatter of the 140 keV ^{99m}Tc photons into the main ^{201}Tl energy window (Kiat et al. 1994, DePuey 1993, Weinstein et al. 1994). Several methods of correcting the crosstalk in simultaneous dual radionuclide scans on standard gamma cameras have been proposed. Moore et al. (1995) uses a simple subtraction of spillover counts based on data measured in two main photopeak windows as well as a third spill-down window. A similar approach has been reported in (Hademenos et al. 1995). A spectral and spatial deconvolution technique, which can be applied to projection data or reconstructed images has been proposed and evaluated in (Knesaurek 1994, Knesaurek & Machac 1997). Hannequin et al. (2000, 2001) describe a method of recovering photopeak counts from projection data collected in multiple subwindows within the usual energy windows. It involves fitting a model with assumed Gaussian energy response of the system to the measured data by linear regression. The potentially most accurate method of crosstalk correction is to model all the contributing physical processes in the simultaneous iterative reconstruction of the two radionuclide distribution images (Kadrmas et al. 1999). Since the analytical treatment of scatter gets very complex beyond the first order, Monte Carlo simulation is usually the most effective way of modelling (Floyd et al. 1986). However, since using direct Monte Carlo simulation in the iterative reconstruction would be very computation intensive and excessively long, various approximations based on precomputed simulation based point spread functions are usually applied (Frey & Tsui 1993, Beekman et al. 2002, Ouyang et al. 2009).

The new dedicated D-SPECT cardiac camera (Spectrum Dynamics, Israel) has several features potentially facilitating simultaneous multiple radionuclide scans: it is based on CdZnTe detectors, which offer much better energy resolution as compared to the standard scintillators, allowing use of narrower energy windows and therefore reducing down-scatter contribution in photopeaks. High sensitivity allows reduction of ^{99m}Tc activity while maintaining standard ^{201}Tl activity so that the relative contribution of crosstalk is decreased. Tungsten collimators have the main X-ray fluorescence peak at about 59 keV, which, combined with high detector energy resolution, allows at least partial separation from the main ^{201}Tl photopeak at 70 keV, as opposed to the 73-75 keV X-ray fluorescence of lead, typical of conventional systems. On the other hand, spectral characteristics of the pixelated solid state detector are considerably different from those of a large crystal scintillator; in particular low energy tails below the main photopeaks are present, due to incomplete charge collection and inter-pixel scatter. Consequently, standard scatter correction methods e.g. based on subtracting scatter counts measured in adjacent narrow windows (Ogawa et al. 1991) are not directly applicable since scatter windows contain significant photopeak spillover.

In this paper we propose a method of extracting photopeak counts from simultaneously acquired ^{99m}Tc and ^{201}Tl projections using multiple energy windows covering the photopeaks, as well as parts of the continuous scatter spectrum and collimator fluorescence X-rays. This is achieved by iteratively solving model equations describing the spatial and spectral response of the system. Rather than trying to model the complex object-dependent scatter and X-ray fluorescence phenomena we make use of the information available in the measured data in order to extract the useful counts. The parameters of the model, which are independent of the scanned object or scanning geometry, are determined from measurements of thin rod sources, when the effects of self-scatter and attenuation can be neglected. Estimation of the parameters can be included as a part of routine system quality control procedures.

2. Methods

2.1. D-SPECT system

D-SPECT is a novel dedicated cardiac SPECT camera with distinctive scanning geometry (Patton et al. 2007, Erlandsson et al. 2009, Slomka et al. 2009, Gambhir et al. 2009). The position adjustable gantry contains 9 pixellated CdZnTe detector modules which can rotate around their axes to acquire region-centric tomographic projections. Scan pattern is adjusted individually to each patient, aiming at maximising the counts from the heart, although projection data of the whole thorax region over 180° is acquired so that there is no truncation problem in the tomographic reconstruction. Detectors are fitted with wide beam tungsten collimators of high sensitivity, enabling significantly reduced scanning times. High image resolution is achieved by applying an iterative reconstruction algorithm involving collimator response modeling.

2.2. D-SPECT in-air and in-patient spectra

To introduce our projection deconvolution method let us begin by examining the spectra measured using sources in air (Fig. 1) when virtually no in-object scatter or attenuation is present. We used the fillable rod source designed for routine system quality control and callibration. A 30 cm long perspex tube of inner diameter 9 mm and wall thickness 2.5 mm was filled with radioactive solutions and placed at a fixed position parallel to detector axes, with distances to detector faces ranging from 19 to 23 cm. Each detector head rotated over 120 angular positions to obtain uniform exposure for each detector point. The spectra shown in Fig. 1 are obtained from all the measured data (all pixels, all positions).

Compared to the spectra obtained on a conventional NaI(Tl) scintillator camera with lead collimator one can notice a significantly better energy resolution of D-SPECT. On the other hand “fat” low energy tails below photopeaks can be observed. These are characteristic for pixellated CdZnTe detectors and are due to incomplete charge collection in the semiconductor as well as incomplete photon detection which may happen, for example when a photon undergoes Compton scatter in one pixel and the scattered photon is stopped in a neighbouring pixel. Tungsten fluorescence peaks at around 59 keV are also visible on the D-SPECT spectrum.

In order to examine the spatial distribution of counts in several energy windows specified in sec. 2.4 a static scan of the same callibration source has been acquired, such that the projection of the rod source was at the centre of each detector head. An example of the ^{201}Tl source profiles at two adjacent energy windows is shown in Fig. 2. The measured FWHM for different energies is plotted versus source-detector distance in Figure 3. It can be seen that the sets of curves are approximately parallel, suggesting that the inter-relationship between resolutions for paired energy windows is independent of distance. We therefore propose that the spatial distributions for energy windows below a photopeak can be obtained, to a first approximation, from the photopeak counts distribution by convolution with a narrow distance-independent Gaussian blurring kernel (whose σ is dependent on the pair of energy windows of interest). The corresponding approximation for the scatter distribution is explored further in section 2.5.

The problem with simultaneous dual $^{99\text{m}}\text{Tc}/^{201}\text{Tl}$ imaging becomes apparent when looking at the spectrum obtained from the anthropomorphic phantom (Fig. 4). There is a significant in-phantom scatter component which spills over into photopeaks, particularly the main ^{201}Tl 70 keV peak.

2.3. Projection counts model

Consider the energy spectrum illustrated in Figure 4; also illustrated are 7 selected energy windows that include the photopeak energy windows at 140 keV for ^{99m}Tc , and at 70 / 167 keV for ^{201}Tl . The additional windows, from which scatter and spillover can be estimated, are selected above and below each photopeak. We consider the measured count distributions, M_i in each energy window i as being formed by a combination of primary photons, P_i , photons scattered in the patient, S_i , and spillover of primary photons into lower energy windows due to collimator-detector effects, T_i ; note that this last term will include K X-ray production in the collimator. This gives $M_i = P_i + S_i + T_i$ but in a scatter window it reduces to $M_i = S_i + T_i$. In order to extract the pure primary photons, P_i , from the measured projections we employ a model of the spectral and spatial count distributions in the selected energy windows. Although the following model can be generalised to any number of energy windows we will limit the description here to the specific case for ^{99m}Tc and ^{201}Tl using the 7 energy windows illustrated in Figure 4.

Our model is developed as follows:

a) The commonly used triple energy window (TEW) scatter correction approach (Ogawa et al. 1991) assumes that the scattered photons in each photopeak energy window i are approximated by the mean of down-scatter from higher energy emissions, as estimated at the upper boundary of the photopeak, $S_{i+1}/\Delta E_{i+1}$, and the total scatter including down-scatter as estimated at the lower photopeak boundary, $S_{i-1}/\Delta E_{i-1}$; we represent this as $S_i = (S_{i+1}/\Delta E_{i+1} + S_{i-1}/\Delta E_{i-1})\Delta E_i/2$, where ΔE_i is the width of energy window i . It is usually assumed that the adjacent windows $i+1$ and $i-1$ are narrow, so that the spatial distributions of scattered counts can be assumed constant between the windows i and $i \pm 1$. Here we modify the TEW approach by allowing broad $i+1$ and $i-1$ windows, and assuming that the spatial distributions of the contributions $S_{i \pm 1}$ to the photopeak window i are related by a convolution with a space-invariant function L_i . Details will be demonstrated later (sec. 2.5) using Monte Carlo simulation.

b) The spatial distribution of spillover counts is related to the photopeak distribution of counts by a convolution with a space-invariant Gaussian function K_{ij} ; note that the spillover

$$T_i = \sum_{j>i} w_{ij} (P_j \otimes K_{ij})$$

will be from all higher energy photopeak windows; i.e. $j \text{ even}$. Parameters were determined by experiment as described later (sec. 2.4).

c) We therefore can define the following equations in which the only unknowns are the primary, P_i , and scatter, S_i , counts in each photopeak energy window.

For i even (photopeak windows):

$$M_i = P_i + \frac{1}{2} \left(S_{i+2} \otimes L_i \frac{\Delta E_i}{\Delta E_{i+2}} + S_i \right) + \sum_{\substack{j>i \\ j \text{ even}}} w_{ij} (P_j \otimes K_{ij}), \quad i=2, 4, 6 \quad (1)$$

For i odd (scatter windows):

$$M_i = S_{i+1} \otimes L_i \frac{\Delta E_i}{\Delta E_{i+1}} + \sum_{\substack{j>1 \\ j \text{ even}}} w_{ij} (P_j \otimes K_{ij}), \quad i=1, 3, 5 \quad (2)$$

Due to the convolution terms, the equations (1,2) for neighbouring pixels are coupled, and therefore the entire set is rather complex. To solve it we use the well known ML-EM algorithm. Here, however, it is applied in the projection space only, so it works much faster than in its more typical application for image reconstruction. Our primary goal is to estimate P_i , the scatter- and crosstalk-free projections. The method is in essence a hybrid between the TEW method and scatter modelling.

2.4. D-SPECT implementation of iterative projection deconvolution

On D-SPECT, data are acquired in list-mode so the counts can be easily sorted into multiple energy windows. We used the following windows:

1. 50 - 64 keV (Tungsten X-rays)
2. 64 - 77 keV (main ^{201}Tl photopeaks)
3. 105 - 130 keV
4. 130 - 149 keV (main $^{99\text{m}}\text{Tc}$ photopeak)
5. 150 - 160 keV
6. 157 - 177 keV (^{201}Tl 167 keV photopeak)
7. 180 - 200 keV

The photopeak windows are non-symmetric with respect to original emission energies, reflecting the fact that CdZnTe spectral response is non-Gaussian (low-energy tails). The value of 64 keV corresponds to the minimum separating the ^{201}Tl main photopeaks and tungsten X-rays (see Fig. 1). The small ^{201}Tl photopeaks around 80 keV were left out due to typically low photopeak-to-scatter ratio in this region. Window 4 includes the weak ^{201}Tl 135 keV peak, which is included in the model as a fixed fraction of the 167 keV peak (parameter w_{46} in Table 1). Window 7 accounts for ^{201}Tl high energy contamination; in most cases it contains only a very low level background and can be discarded.

The coefficients w_{ij} as well as the parameters of the Gaussian convolution kernels K_{ij} in the equations (1,2) were determined by examining scans of thin rod sources (no in-object scatter) in the selected energy windows (cf. sec. 2.2). The spillover parameters w_{ij} were calculated by taking the ratios of total numbers of counts in respective energy windows acquired in the uniform scan of the $^{99\text{m}}\text{Tc}$ and ^{201}Tl rod sources over all detector pixels. Additionally, in order to estimate the spillover from the 167 keV ^{201}Tl photopeak to low energy windows 1 and 2, which are overshadowed by strong low energy emissions in ^{201}Tl spectrum, we scanned a ^{123}I source with its 159 keV emission. The energy windows have been rescaled appropriately to obtain the parameters w_{16} and w_{26} .

In order to compute the parameters of the Gaussian convolution kernels K_{ij} we used the static scans of rod sources at distances 19 - 23 cm to detector faces. The transaxial profiles of the projections were fitted with Gaussian functions for all energy windows and source - detector distances (cf. Fig 2,3). The respective standard deviations were computed as

$\sigma_{ij} = \sqrt{\sigma_i^2 - \sigma_j^2}$, averaged over all the measured distances. The values of the parameters are given in table 1.

Since the convolution kernels K_{ij} describe the detector and collimator effects, they are constrained to the size of detector heads, 4 cm $\times$ 16 cm. This is in contrast to the scatter convolution kernels L_i as will be discussed in the next section. Detector edge effects are included in appropriate pixel sensitivity maps.

2.5. Spatial distribution of scattered photon projections

It is well known that the distribution of photons emitted from a point source and scattered in an object (patient) can be well approximated by monoexponential functions on the planar projections (Floyd et al. 1985, Kadrmas et al. 1997). They are, in general, anisotropic and the parameters of the distribution in each direction depend on the scattering object and the position of the source in it. It can be shown (Meikle et al. 1991) that if the distribution of a point source scatter projection in a photopeak energy window, $i + 1$, can be modelled as a convolution of the primary photons with a mono-exponential function: $S_{i+1} = P_{i+1} \otimes A_{i+1} \exp(-\lambda_{i+1}r)$ and similarly in the window i below: $S_i = P_i \otimes A_i \exp(-\lambda_i r)$ then the respective scatter projection distributions are related by a convolution with the kernel L_i :

$$S_i = S_{i+1} \otimes C_i (\delta + \alpha_i \exp(-\lambda_i r)) \equiv S_{i+1} \otimes L_i, \quad (3)$$

where $C_i = A_i/A_{i+1}$, $\alpha_i = \lambda_{i+1} - \lambda_i$ and δ is a Dirac delta. In other words, the scatter distribution in the lower window is a weighted sum of the scatter projection in the upper window and a convolution of that projection with a mono-exponential. The relationship between scatter distributions measured in two neighbouring energy windows demonstrates less depth dependence than the scatter distribution itself. To show this we performed Monte Carlo simulations of ^{99m}Tc and ^{201}Tl point sources located on the axis of a water filled cylindrical phantom of length 60 cm and 60 cm in diameter, using the SIMIND software package (Ljungberg & Strand 1989). A continuous detector with the energy resolution of CZT fitted with a tungsten collimator of hole size as that of the D-SPECT was used. Detector and collimator effects such as scattering and secondary X-ray emission were taken into account. We generated planar scatter projections on a detector at the base of the cylinder for the sources at a range of distances: 5, 10, 15, 20 and 30 cm from the detector. The data have been sorted into the energy windows specified above. For each position of the source and each pair of energy windows there is a unique set of parameters α_i , λ_i , C_i which determines the relation between scatter distributions in these windows by means of Eq. (3). However, in order to find an approximate position-independent relationship for the window pairs of interest, we scanned the (α_i, λ_i) parameter space and found a single pair of parameters which minimises the sum of mean square errors between the fit with Eq. (3) and the measured distributions for all source positions. The parameter C_i which does not affect the shape of the distribution, but just the normalisation, has been ignored. The resulting parameters defining the scatter convolution kernels are shown in Table 2.

We checked that the approximate position-independent convolution kernels determined in this way give a good approximation of the true relationship between scatter distributions in different energy windows. Two examples are shown in Figures 5 and 6. Discrepancies can be noticed mainly for deeply located sources, which in routine practice add only small contributions to the total projection counts due to long attenuation paths.

Our assumption has also been verified by Monte Carlo simulations of a numerical NCAT phantom (Segars et al. 1999) using the SIMIND software package (Ljungberg & Strand

1989). We used a ^{99m}Tc activity distribution typical for myocardial perfusion studies and simulated projection data in the photopeak window 130 - 149 keV and the scatter window 105 - 130 keV, recording in-patient scattered photons only. The scattered counts distribution in the photopeak has been convolved with the mono-exponential kernel L_3 and compared with the distribution actually measured in the scatter window (Fig.7). Good agreement of both distributions was observed.

As explained in sec. 2.1 in D-SPECT projections are acquired by 9 small detector heads rather than a single flat panel detector. In contrast to the convolution kernels K_{ij} which reflect the detector and collimator effects, the scatter kernels L_j describe the in-patient scatter influenced by activity outside each detector's geometric field-of-view. As a consequence, in order to compute convolutions for an extended field-of-view prior to truncation we need to rebin the measured data to obtain conventional planar projections, then perform the convolution, and finally apply an inverse rebinning to go back to detector head projections.

2.6. ML-EM implementation

Equations (1) and (2) describe the linear problem to be solved by the ML-EM algorithm, with right-hand sides defining the system matrix. It is highly sparse and nearly diagonal. The problem is solved simultaneously for all P_j and S_j . The numbers of ^{201}Tl photopeak counts in the windows 64-77 keV and 157-177 keV are treated as two independent unknown variables. The final ^{201}Tl image is formed as a sum of the two solutions.

In order to control noise in the solution we applied inter-iteration spatial smoothing of the projection estimates P_j and S_j with Gaussian kernels. All the terms in Eqs. (1) and (2) have also been corrected by appropriate pixel sensitivity maps.

2.7. Phantom experiments

To evaluate the method with measured data we performed scans of a thorax phantom (Data Spectrum, Hillsborough, NC, USA) with a cardiac insert containing two fillable defects. The phantom was filled with a mixture of ^{99m}Tc and ^{201}Tl in concentrations similar to typical clinical values. The defects contained reduced activities of one of the radionuclides while maintaining the concentration of the other at the same level as in the normal myocardium. Two phantom configurations were scanned:

1. small (45° around LV) 100% defects
 activity in LV: 14.4 MBq ^{99m}Tc + 6.5 MBq ^{201}Tl
 activity in background: 30 MBq ^{99m}Tc + 75 MBq ^{201}Tl
2. large (^{99m}Tc : 90° , ^{201}Tl : 180° around LV) 70% defects
 activity in LV: 9.5 MBq ^{99m}Tc + 7.3 MBq ^{201}Tl
 activity in background: 14.5 MBq ^{99m}Tc + 42 MBq ^{201}Tl
 activity in liver: 11.5 MBq ^{99m}Tc + 48 MBq ^{201}Tl
 activity in bottom cylinder: 53 MBq ^{99m}Tc + 33 MBq ^{201}Tl

No activity was put in the lung compartments. The second configuration models extra-cardiac activity more realistically. The bottom cylinder, on which the anthropomorphic phantom was standing, modeled the bowel activity. In the absence of an actual liver compartment in the phantom, the liver was represented by a 500 mL saline bag attached to the bottom of the phantom. The measured projections were deconvolved using the method described above. The resulting pure photopeak projections were then reconstructed with the

standard Ordered Subsets Expectation Maximisation (OS-EM) algorithm with collimator resolution modeling and no attenuation correction. ^{201}Tl images were reconstructed using the sums of counts in windows 2 and 6 for the raw data and, respectively, the sums $P_2 + P_6$ for deconvolved projections. Using the registered segmented CT scan we calculated lesion contrast defined as $C = (I_M - I_D)/I_M$, where I_M and I_D are the average intensities in the myocardium and the defect, respectively.

In order to assess the sensitivity of the iterative deconvolution of projections to noise, 10 consecutive 4 minute scans of the phantom with large 70% defects (configuration 2) were acquired. The projections were processed with the width of inter-iterative Gaussian smoothing kernels varying from $\sigma = 0$ to $\sigma = 2$ pixels. The respective contrasts and average coefficients of variation over the LV wall were calculated to produce contrast noise curves.

2.8. Patient studies

The proposed method has been applied to several studies of patients undergoing simultaneous ^{201}Tl rest/ $^{99\text{m}}\text{Tc}$ stress scans. The patients were injected with about 80 MBq of ^{201}Tl at rest and about 30 minutes later scanned on D-SPECT for 6 minutes. Subsequently, 250 MBq $^{99\text{m}}\text{Tc}$ sestamibi was injected during adenosine infusion (pharmacological stress). Then a 15 minute D-SPECT simultaneous dual radionuclide scan followed about 75 minutes after stress injection. The raw and corrected data were reconstructed using the standard clinical OS-EM algorithm. The first ^{201}Tl -only scan was used as a reference for the ^{201}Tl rest image obtained from the second dual radionuclide scan.

3. Results

3.1. Phantom experiments

The phantom with small 100% defects (configuration 1) was scanned for 6 minutes. The same phantom was scanned again 3 days later to obtain a ^{201}Tl only scan with virtually no $^{99\text{m}}\text{Tc}$ for comparison. Snapshots of the resulting images reconstructed from raw and processed projections are shown in Fig. 8.

Without correction the ^{201}Tl defect appears with poor contrast, while after the iterative deconvolution processing it becomes clearly evident. The contrast in the processed ^{201}Tl image improves by 50%, while there is only marginal improvement in the $^{99\text{m}}\text{Tc}$ image which contains little crosstalk and scatter even without processing. The numerical values of defect contrasts are given in Table 3.

The iterative deconvolution also improves the contrast of the ^{201}Tl only scan, which indicates that it could form the basis for an efficient scatter correction method when a single radionuclide is used.

Fig. 9 shows the contrast-noise curves obtained for the configuration 2 phantom. Note that the choice of parameters for optimal compromise between contrast and noise is related to the clinical application. For ^{201}Tl images the inter-iterative smoothing kernel with $\sigma = 1$ yields the best performance, significantly improving true defect contrast and reducing crosstalk from the $^{99\text{m}}\text{Tc}$ defect, while keeping noise at an acceptable level. As in the previous case, the projection deconvolution does not noticeably alter the $^{99\text{m}}\text{Tc}$ images. There is a difference in the optimal numbers of iterations for ^{201}Tl and $^{99\text{m}}\text{Tc}$ activities. The true contrast for ^{201}Tl increases with the number of iterations, while the false contrast decreases initially and is similar at 5 and 10 iterations for $\sigma = 1$. For $^{99\text{m}}\text{Tc}$ on the other hand, the true contrast does not increase significantly between 5 and 10 iterations, while the false contrast does increase. Therefore 10 iterations can be used for ^{201}Tl while 5 is better for $^{99\text{m}}\text{Tc}$. This is consistent with the fact that the main ^{201}Tl photopeak window is more contaminated by

downscatter and cross-talk counts so it takes longer for the algorithm to converge. Fig. 10 shows polar plots of the left ventricle with large defect reconstructed from raw data and deconvolved with different number of iterations. The defect contrast clearly increases with the deconvolution. However, there is also a false cold area in the basal-inferior wall. The reason for this is unclear.

3.2. Patient studies

The images reconstructed from the data processed by our correction algorithm showed reduced scatter related background and improved contrast compared to non-corrected dual radionuclide studies in the majority of cases, consistent with the observation in phantom studies. A single representative study is shown in Fig 11. The uncorrected images indicate a partially reversible defect in the infero-lateral wall. After correction the defect in the ^{201}Tl rest image is more pronounced suggesting a smaller degree of reversibility. A more complete analysis of the initial group of patients was recently published (Ben-Haim et al. 2010). In this paper perfusion images obtained using dual radionuclide acquisition were compared with those obtained using conventional acquisition and diagnostic interpretation was demonstrated to be comparable. For summed stress and rest scores, high correlation was demonstrated between conventional and dual radionuclide SPECT. The dual radionuclide studies therefore demonstrate similar clinical utility with the advantage of requiring significantly reduced study time. Further clinical evaluation is ongoing.

4. Discussion

We have developed a method of deconvolution of simultaneously acquired $^{99\text{m}}\text{Tc}$ and ^{201}Tl data based on a spatial and spectral model. The method involves a number of approximations. The main one relates to the in-patient scattered photons distribution, which can only be properly modelled in the object space with a known attenuation map and estimated activity distribution (Kadmas et al. 1999). In our projection based approach we assume that the scattered photons detected in each photopeak window and associated “scatter window” is related by convolution with an object-independent mono-exponential kernel. Our simulations confirmed that this assumption is reasonable for typical geometry of cardiac scans. To assess the amount of scatter in the photopeak window we assume, following the TEW method, that it is equal to the average of corresponding scatter amplitudes in neighbouring energy windows. In fact, careful examination of phantom and patients’ data may provide a better approximation by fitting a linear combination of the amplitudes rather than simply the average.

The model equations (1) and (2) describe the linear “forward” problem which can be solved by various methods. In all examples shown in this paper we used the ML-EM algorithm applied to all equations simultaneously. Another possibility is to use the fact that high energy windows are not affected by counts in lower energy windows, so the Eqs. (1) and (2) could be solved sequentially “two by two” starting from the high energies and using the solutions as input to lower energy windows. This will be explored in future studies.

As a result of our deconvolution algorithm we obtain the estimates of pure ^{201}Tl and $^{99\text{m}}\text{Tc}$ photopeak count projections which obviously do not have Poisson noise statistics. It is therefore inappropriate to apply the conventional ML-EM algorithm for subsequent image reconstruction. A more appropriate approach would be to include our projection model in the reconstruction projector. In this way the reconstruction and deconvolution iterations would be performed simultaneously. Such implementation, however, would require modification of the current D-SPECT reconstruction algorithm and would be less efficient

than running deconvolution as a separate pre-processing step. It was more convenient in this initial investigation to extract scatter-free projections as a separate pre-processing step.

5. Conclusions

We demonstrated that iterative deconvolution of measured projections significantly improves the quality of images obtained during simultaneous dual radionuclide ^{99m}Tc and ^{201}Tl scans by reducing crosstalk and scatter effects, while maintaining acceptable noise level. It is fast, independent of the image reconstruction algorithm, and uses information from multiple energy windows in addition to the photopeak windows. Although the method has been developed for a novel dedicated cardiac SPECT camera based on CdZnTe pixellated detectors, it can also be adapted to any conventional scintillator camera. Further refinement of the method is possible in terms of optimising the choice of energy windows or the in-patient scatter model. A similar approach can also be applied to other pairs of radionuclides e.g. ^{99m}Tc and ^{123}I , or as a scatter correction method for single radionuclide studies.

Acknowledgments

We acknowledge helpful discussions with Nathaniel Roth and other staff of Spectrum Dynamics. This work has been partly supported by the Wellcome Trust grant no 084288/Z/07/Z and CRUK/EPSRC Comprehensive Cancer Imaging Centre funding. The Institute of Nuclear Medicine receives a proportion of its funding from the Department of Health NIHR Biomedical Research Centre funding scheme.

References

- Beekman FJ, de Jong HW, van Geloven S. Efficient fully 3-D iterative SPECT reconstruction with Monte Carlo-based scatter compensation. *IEEE Trans Med Imaging*. 2002; 21(8):867–877. [PubMed: 12472260]
- Ben-Haim S, Kacperski K, Hain S, Van Gramberg D, Hutton BF, Erlandsson K, Sharir T, Roth N, Waddington WA, Berman DS, Ell PJ. Simultaneous dual-radionuclide myocardial perfusion imaging with a solid-state dedicated cardiac camera. *Eur J Nucl Med Mol Imaging*. 2010; 37(9): 1710–1721. [PubMed: 20383705]
- Berman DS, Kiat H, Van Train K, Friedman JD, Wang FP, Germano G. Dual-isotope myocardial perfusion spect with rest thallium-201 and stress tc-99m sestamibi. *Cardiol Clin*. 1994; 12(2):261–270. [PubMed: 8033176]
- DePuey EG. Simultaneous thallium-201/technetium-99m dual-isotope cardiac SPECT: ready for prime time? *J Nucl Med*. 1993; 34(11):2006–2008. [PubMed: 8229251]
- Erlandsson K, Kacperski K, van Gramberg D, Hutton BF. Performance evaluation of D-SPECT: a novel SPECT system for nuclear cardiology. *Phys Med Biol*. 2009; 54(9):2635–2649. [PubMed: 19351981]
- Floyd CEJ, Jaszcak RJ, Greer KL, Coleman RE. Deconvolution of compton scatter in SPECT. *J Nucl Med*. 1985; 26(4):403–408. [PubMed: 3872353]
- Floyd CE, Jaszcak RJ, Greer KL, Coleman RE. Inverse monte carlo as a unified reconstruction algorithm for ECT. *J Nucl Med*. 1986; 27(10):1577–1585. [PubMed: 3489822]
- Frey EC, Tsui BMW. A practical method for incorporating scatter in a projector-backprojector for accurate scatter compensation in SPECT. *IEEE Transactions on Nuclear Science*. 1993; 40:1107–1116.
- Gambhir SS, Berman DS, Ziffer J, Nagler M, Sandler M, Patton J, Hutton B, Sharir T, Haim SB, Haim SB. A novel high-sensitivity rapid-acquisition single-photon cardiac imaging camera. *J Nucl Med*. 2009; 50(4):635–643. [PubMed: 19339672]
- Hademenos GJ, Dahlbom M, Hoffman EJ. Simultaneous dual-isotope technetium-99m/thallium-201 cardiac SPET imaging using a projection-dependent spill-down correction factor. *Eur J Nucl Med*. 1995; 22(5):465–472. [PubMed: 7641755]

- Hannequin P, Mas J, Germano G. Photon energy recovery for crosstalk correction in simultaneous $^{99m}\text{Tc}/^{201}\text{Tl}$ imaging. *J Nucl Med*. 2000; 41(4):728–736. [PubMed: 10768576]
- Kadrmas DJ, Frey EC, Tsui BM. Analysis of the reconstructibility and noise properties of scattered photons in Tc-^{99m} SPECT. *Phys Med Biol*. 1997; 42:2493–2516. [PubMed: 9434303]
- Kadrmas DJ, Frey EC, Tsui BM. Simultaneous technetium- 99m /thallium- 201 SPECT imaging with model-based compensation for cross-contaminating effects. *Phys Med Biol*. 1999; 44(7):1843–1860. [PubMed: 10442716]
- Kiat H, Germano G, Friedman J, Van Train K, Silagan G, Wang FP, Maddahi J, Berman D. Comparative feasibility of separate or simultaneous rest thallium- 201 /stress technetium- 99m -sestamibi dual-isotope myocardial perfusion SPECT. *J Nucl Med*. 1994; 35(4):542–548. [PubMed: 8151372]
- Knesaurek K. A new dual-isotope convolution cross-talk correction method: a $\text{Tl-}^{201}/\text{Tc-}^{99m}$ SPECT cardiac phantom study. *Med Phys*. 1994; 21(10):1577–1583. [PubMed: 7869989]
- Knesaurek K, Machac J. Enhanced cross-talk correction technique for simultaneous dual-isotope imaging: a $\text{Tl-}^{201}/\text{Tc-}^{99m}$ myocardial perfusion SPECT dog study. *Med Phys*. 1997; 24(12):1914–1923. [PubMed: 9434974]
- Ljungberg M, Strand SE. A Monte Carlo program for the simulation of scintillation camera characteristics. *Comput Methods Programs Biomed*. 1989; 29(4):257–272. [PubMed: 2791527]
- Meikle, SR.; Hutton, BF.; Bailey, DL.; Fulton, RR.; Schindhelm, K. Information Processing in Medical Imaging. Colchester, ACF.; Hawkes, DS., editors. Springer Verlag; Berlin: 1991. p. 34–44.
- Moore SC, English RJ, Syravanh C, Tow DE, Zimmerman RE, Chan KH, Kijewski MF. Simultaneous $\text{Tc-}^{99m}/\text{Tl-}^{201}$ imaging using energy-based estimation of the spatial distributions of contaminant photons. *IEEE Trans Nucl Sci*. 1995; 42(4):1189–1195.
- Ogawa K, Harata Y, Ichihara T, Kubo A, Hashimoto S. A practical method for position-dependent Compton-scatter correction in single photon emission CT. *IEEE Trans Med Imaging*. 1991; 10(3):408–412. [PubMed: 18222843]
- Ouyang J, Zhu X, Trott CM, El Fakhri G. Quantitative simultaneous $^{99m}\text{Tc}/^{123}\text{I}$ cardiac spect using mc-josem. *Med Phys*. 2009; 36(2):602–611. [PubMed: 19292000]
- Patton JA, Slomka PJ, Germano G, Berman DS. Recent technologic advances in nuclear cardiology. *J Nucl Cardiol*. 2007; 14(4):501–513. [PubMed: 17679058]
- Segars WP, Lalush DS, Tsui BMW. A realistic spline-based dynamic heart phantom. *IEEE Trans Nucl Sci*. 1999; 46(3):503–506.
- Slomka PJ, Patton JA, Berman DS, Germano G. Advances in technical aspects of myocardial perfusion SPECT imaging. *J Nucl Cardiol*. 2009; 16(2):255–276. [PubMed: 19242769]
- Weinstein H, King MA, Reinhardt CP, McSherry BA, Leppo JA. A method of simultaneous dual-radionuclide cardiac imaging with technetium- 99m and thallium- 201 . I: Analysis of interradionuclide crossover and validation in phantoms. *J Nucl Cardiol*. 1994; 1(1):39–51. [PubMed: 9420669]

Figure 1. Spectra of uniformly scanned thin rod ^{99m}Tc and ^{201}Tl sources measured in air for D-SPECT and conventional gamma camera (Prism 3000, Philips, Cleveland, OH)

Figure 2.

Count profiles through the static projections of a ^{201}Tl rod source in air on a single detector head measured in the main photopeak window 64-77 keV and the lower window 50-64 keV containing mainly tungsten fluorescence X-rays. Solid lines represent Gaussians fitted to measured data points.

Figure 3.
The width (FWHM) of ^{201}Tl rod source static projections as a function of distance to the detector for the specified energy windows.

Figure 4. Total spectrum of an anthropomorphic phantom filled with $^{99\text{m}}\text{Tc}$ and ^{201}Tl activities as described in section 2.7 (configuration 1), measured on D-SPECT. The energy windows described in sec. 2.4 are marked.

Figure 5. Simulated distribution of scattered photons for a point ^{99m}Tc source in a cylinder 60×60 cm at varying depths. Dotted lines - scatter in the 130-149 keV photopeak window, solid lines - true scatter in the 105-130 keV window, dotted-dashed lines - estimations of scatter in the 105-130 keV window using Eq. 3 with fixed parameters $\alpha_3 = 0.43$ and $\lambda_3 = 0.18 \text{ cm}^{-1}$.

Figure 6. Simulated distribution of scattered photons for a point ^{201}Tl source in a cylinder 60×60 cm at varying depths. Dotted lines - scatter in the 64-77 keV photopeak window, solid lines - true scatter in the 50-64 keV window, dotted-dashed lines - estimations of scatter in the 50-64 keV window using Eq. 3 with fixed parameters $\alpha_1 = 0.22$ and $\lambda_1 = 0.2 \text{ cm}^{-1}$.

Figure 7.

A profile through heart projection of scattered counts in the scatter energy window (105-130 keV) generated from the NCAT phantom using Monte Carlo simulation (points) and an approximated scatter distribution obtained by convolving the scatter projection in photopeak window (130-149 keV) with the kernel given by Eq. (3) with $\alpha_3 = 0.43$ and $\lambda_3 = 0.18 \text{ cm}^{-1}$ (solid line).

Figure 8.

Reconstructed images of the phantom with small 100% defects (6 minute scan): short axis slices, horizontal long axis slices and polar map for reconstructions based on raw (windows 2+6 and 4) and deconvolved projections; inter-iterative smoothing with a Gaussian kernel with $\sigma = 1$; ^{99m}Tc images - 5 iterations, ^{201}Tl images - 10 iterations.

Figure 9.

Contrast noise curves for the phantom with large 70% defects. Top graph shows measured true defect contrast (in perfect image it should be 0.7); bottom graph shows contrast in defect regions for non reduced activity (should be 0 in perfect image). Points on each curve correspond to 1, 2, 3, 5, and 10 iterations. Larger false defect contrast in ^{99m}Tc images is mainly due to attenuation effect in the inferior wall.

Figure 10.

Polar plots of LV ^{201}Tl images of the phantom with large 70% defects (4 minute dual radionuclide scan) reconstructed from raw data ([64 - 77 keV] + [157 - 177 keV] windows), and projections deconvolved with inter-iterative smoothing (Gaussian kernel with $\sigma = 1$). Images obtained from ^{201}Tl -only scan of the same phantom are shown for comparison. Activity concentrations (perfusion) calculated by the QPS software.

Figure 11.

Simultaneous ^{201}Tl rest - ^{99m}Tc stress study of a 73y old male: (a) uncorrected data reconstruction for 20% 140 keV and 15% 70 keV energy windows; (b) reconstruction from data processed by the iterative deconvolution algorithm; (c) rest image obtained from uncorrected ^{201}Tl only scan (prior to injection of ^{99m}Tc). The arrows indicate the defect in infero-lateral wall, enhanced after crosstalk/scatter correction.

Table 1

Parameter values in the projection counts model, Eqs. (1,2), for the set of energy windows specified in sec. 2.4.

spillover weights		blur kernels STD	
parameter	value	parameter	value [mm]
w_{12}	0.343	K_{12}	6.77
w_{14}	0.140	K_{14}	11.53
w_{24}	0.105	K_{24}	10.53
w_{34}	0.447	K_{34}	5.58
w_{16}	0.278	K_{16}	11.53
w_{26}	0.185	K_{26}	10.53
w_{36}	0.520	K_{36}	6.77
w_{46}	0.585	K_{46}	4.23
w_{56}	0.288	K_{56}	5.73

Table 2

Parameter values of the scatter convolution kernels, Eq. (3), for the energy windows specified in sec. 2.4.

Energy windows [keV]		scatter blur kernels L_i [Eq. (3)]			Isotope used
photopeak	lower	parameter	α_i	λ_i [cm ⁻¹]	
157 – 177	150 – 160	L_5	0.11	0.065	²⁰¹ Tl
157 – 177	130 – 149	L_4	0.39	0.22	²⁰¹ Tl
130 – 149	105 – 130	L_3	0.43	0.18	^{99m} Tc
130 – 149	64 – 77	L_2	2.7	0.165	^{99m} Tc
64 – 77	50 – 64	L_1	0.22	0.20	²⁰¹ Tl + ^{99m} Tc

Table 3

Defect contrast in raw and processed images of cardiac phantom with small 100% defects. For a perfect image it should be equal to 1. Iterative deconvolution with 5 iterations for ^{99m}Tc images and 10 iterations for ^{201}Tl images; inter-iterative smoothing with a Gaussian kernel with $\sigma = 1$.

	Defect contrast		
	^{201}Tl	^{99m}Tc	^{201}Tl only
Raw data	0.26	0.41	0.38
Iterative deconvolution	0.39	0.43	0.42