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Performance evaluation of D-SPECT: a novel SPECT system for nuclear cardiology

Kjell Erlandsson1, Krzysztof Kacperski, Dean van Gramberg and Brian F Hutton

Institute of Nuclear Medicine, University College London and UCLH NHS Foundation Trust, London NW1 2BU, UK

E-mail: kjell.erlandsson@uclh.nhs.uk

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Abstract
D-SPECT (Spectrum Dynamics, Israel) is a novel SPECT system for cardiac perfusion studies. Based on CZT detectors, region-centric scanning, high-sensitivity collimators and resolution recovery, it offers potential advantages over conventional systems. A series of measurements were made on a β-version D-SPECT system in order to evaluate its performance in terms of energy resolution, scatter fraction, sensitivity, count rate capability and resolution. Corresponding measurements were also done on a conventional SPECT system (CS) for comparison. The energy resolution of the D-SPECT system at 140 keV was 5.5% (CS: 9.25%), the scatter fraction 30% (CS: 34%), the planar sensitivity 398 s MBq−1 per head (99mTc, 10 cm) (CS: 72 s MBq−1), and the tomographic sensitivity in the heart region was in the range 647–1107 s MBq−1 (CS: 141 s MBq−1). The count rate increased linearly with increasing activity up to 1.44 M s−1. The intrinsic resolution was equal to the pixel size, 2.46 mm (CS: 3.8 mm). The average reconstructed resolution using the standard clinical filter was 12.5 mm (CS: 13.7 mm). The D-SPECT has superior sensitivity to that of a conventional system with similar spatial resolution. It also has excellent energy resolution and count rate characteristics, which should prove useful in dynamic and dual radionuclide studies.

(Some figures in this article are in colour only in the electronic version)

1. Introduction

The D-SPECT system (Spectrum Dynamics, Caesarea, Israel), based on novel detector technology and a unique acquisition geometry, offers potential advantages in nuclear cardiology compared to conventional gamma camera single photon emission computed...
Figure 1. Schematic drawing of the D-SPECT scanner, indicating the position of the nine detectors in the gantry. A SPECT/CT image is also included, representing a patient being scanned.

tomography (SPECT) systems. For many years, the design of SPECT systems has been dominated by the conventional Anger camera (Anger 1957, 1964). The primary components of this system are a collimator that limits the direction of incoming detectable photons and a scintillation detector from which the location and energy of photons are determined by means of a set of photomultiplier tubes optically coupled to the detector. The sensitivity of this system is largely limited by constraints in the physical collimation. The appeal of the conventional gamma camera has been its flexibility in providing SPECT capability while preserving the ability to perform general planar studies. The geometry of acquisition usually involves rotation of the detector around the patient, although dedicated SPECT systems have been designed by positioning a fixed array of detectors around the patient (Rogers et al 1982). Most commonly for cardiac acquisition dual detectors are used where both speed of rotation and detector location tend to be limited by the bulk of the overall system.

The disadvantages of scintillation detectors include bulkiness and relatively poor energy resolution. Solid-state detectors have long been used in spectroscopic applications due to their superior energy resolution, but have not been widely used in medical imaging for reasons of practicality and cost. Solid-state detectors based on cadmium zinc telluride (CZT) can operate at room temperature, and recent technical advances have led to the development of pixelated CZT detector units appropriate for medical imaging applications (Wagenaar 2004). The absence of PM tubes allows for a compact and flexible design.

The global demand for SPECT cardiac perfusion studies has resulted in SPECT instruments being developed that are dedicated to this purpose, whether using compact conventional cameras or alternative designs (Chang et al 2006, Patton et al 2007). The D-SPECT has been specifically designed for this market. It consists of nine arrays of CZT detectors, each with a dimension of approximately 4 cm × 16 cm. Each detector rotates around its central axis with programmable angular orientation (see figure 1). The compact nature of the CZT detector permits movement that would not be achievable with conventional camera designs. The system therefore permits ‘region-centric’ acquisition where it is programmed to acquire data mainly from a pre-selected region that includes the heart, maximizing the acquired counts from this region. The detectors are equipped with tungsten collimators that have larger aperture area and shorter length than conventional collimators, providing a high sensitivity although with relatively poor geometric resolution. Central to the system is an iterative reconstruction algorithm that permits the restoration of resolution by including a
model of the collimator geometry within the system model. A more detailed description of the system is provided elsewhere (Patton et al 2007, Gambhir et al 2009). Initial clinical results from the system have been recently reported (Sharir et al 2008).

The potential advantages of the D-SPECT compared to a conventional SPECT system are better energy resolution due to the use of CZT and higher sensitivity due to the use of wide-angle collimators combined with region-centric data acquisition. The spatial resolution should be good due to the use of resolution recovery during reconstruction. Our objective in this paper is to describe a set of measurements that have been carried out to characterize the performance of this novel system. The D-SPECT performance was evaluated in terms of energy resolution, scatter fraction, detector sensitivity, tomographic sensitivity, count rate capability, intrinsic resolution and reconstructed resolution.

2. Materials and methods

2.1. Scanners

This evaluation was done using a β-version D-SPECT scanner placed at the Institute of Nuclear Medicine, London, UK. The scanner consists of nine blocks with pixelated CZT detectors, where each detector block is composed of $16 \times 64$ individual pixels with a spacing of 2.46 mm in both dimensions, resulting in a total detector surface of 39 mm $\times$ 157 mm.

During data acquisition in the standard scanning mode, each of the nine individual detector blocks rotates around its central axis in order to cover the whole field of view (FOV). The detector blocks are also translated in order to obtain a complete tomographic sampling (Gambhir et al 2009). This type of acquisition is called sweep mode.

At the beginning of each study, a quick scout scan is performed (typically for 30 s) in which each detector block sweeps uniformly across the whole FOV, and a preliminary image of the activity distribution is obtained. The operator defines an ROI that should include the whole heart, but not much else. This ROI is then used to generate a specific scanning pattern, designed in such a way that each detector should spend more time acquiring data coming from the direction of the ROI than from regions outside. Data from the whole FOV are still acquired, however, so as to avoid truncation. Any pattern of detector head motion can be set up including the static acquisition mode, i.e. with motionless detectors (although not part of the clinical procedures).

Reconstruction is performed using a variant of the OSEM algorithm (Hudson and Larkin 1994) including modeling of distance-dependent resolution during reconstruction, but no correction for attenuation and scatter. The reconstruction process consists of two stages, each one consisting of a number of OSEM iterations ($N$). After the first stage, a spatial prior, corresponding to the left ventricular wall, is applied to define a starting image for the second reconstruction stage. Inter-iteration filtering is applied by convolution with a smoothing kernel with a central value of $(1 - w)$. A nonlinear post-filter is applied at the end to remove hot-spots. For further details, see Gambhir et al (2009).

In this work, we used the standard reconstruction parameters, recommended for clinical use, except where indicated. The standard parameters were: for the first stage: $N = 3$, $w = 0.2$, and for the second stage: $N = 4$, $w = 0.125$. The number of subsets used in OSEM was 32. The voxels of the reconstructed images are cubic with sides of 4.92 mm.

For comparison, we also performed some scans on a conventional dual-headed SPECT system, the Infinia (GE Healthcare, Haifa, Israel), using standard acquisition parameters for cardiac studies (table 1). Energy resolution was measured on a Prism 3000 system (Philips Medical Systems, Cleveland, USA).
2.2. Energy resolution

One advantage of CZT detectors as compared with NaI(Tl) detectors is superior energy resolution. Energy spectra were obtained for five different radionuclides: $^{57}$Co, $^{99m}$Tc, $^{123}$I, $^{153}$Gd and $^{201}$Tl. In two cases ($^{57}$Co and $^{153}$Gd), sealed rod sources (diameter approximately 1 mm, height 180 mm) were used. In the other cases, a fillable cylindrical source (diameter 10 mm, height 275 mm) was used. The sources were placed vertically at approximately equal distance (about 200 mm) from all detectors. Data were acquired in the sweep mode, i.e. each detector module rotated around its own axis, uniformly exposing all pixels. Energy spectra were generated for each pixel using individual calibration parameters, and an average spectrum was then obtained across all pixels in all detectors. The energy linearity was investigated by looking at the positions of the five main peaks.

The energy resolution (FWHM) was estimated for each pixel separately by fitting the sum of a Gaussian and a linear function, reporting the Gaussian FWHM. The resulting FWHM values were then averaged across pixels. For the composite $^{201}$Tl peaks, a simple correction was made in order to take into account the multiple photon energies. The correction was based on the assumption that the peak was composed of 37% 68.9 keV photons ($K_{a2}$) and 63% 70.8 keV photons ($K_{a1}$). Convolution of these mono-energetic lines with a known Gaussian function and estimation of the width of the resulting curve yielded a correction term that was applied to the data. The uncorrected FWHM values, measured directly from the peaks in the mean energy spectra without background subtraction, are also reported.

Energy spectra were also obtained from the Prism 3000 scanner and the FWHM of the peaks were estimated as described above.

2.3. Scatter fraction

The reconstructed scatter fraction was obtained from a scatter distribution profile. A $^{99m}$Tc rod source ($A = 180$ MBq) was placed at the centre of a 20 cm × 20 cm cylindrical phantom filled with water. Data were acquired on the D-SPECT for 20 min in the sweep mode with an energy window of 20%. Transverse reconstructed image planes were summed over a 100 mm range centrally along the rod source. A 2D mono-exponential function, centred at the source position, was fitted to the image values within a range of 40–80 mm from the source. A 2D scatter distribution was generated by extrapolating the fitted function back to the source position and out to a distance of 100 mm from the source. The reconstructed scatter fraction ($\text{SF}_{\text{rec}}$) was calculated as the integral under the scatter distribution divided by the integral of the reconstructed image within a radius of 100 mm from the source.
On the Infinia a single projection was acquired over 20 min. A tomographic image, reconstructed assuming rotational symmetry, was analysed as described above.

2.4. Detector sensitivity

The D-SPECT system demonstrates a significant improvement in sensitivity compared to conventional systems, due to the combined effect of wide-angle collimation and region-centric acquisition. Here we measured the sensitivity of one single detector. The absolute sensitivity was measured using a point source (a 3 mm × 3 mm cylindrical cavity) containing a solution of 99mTc. The activity in the source was determined by measuring the activity in the syringe used for filling the source in a calibrated well counter (CRC-15R, Capintec, Ramsey, USA), before and after filling the source. The point source was placed at 10 cm from one of the detector arrays (detector #0), and data were acquired for 1 min in the static mode. The total number of counts obtained in a 20% energy window centred at the 140 keV peak was recorded. After removing the point source, a 5 min background measurement was performed. The system sensitivity was determined as the background-subtracted count rate divided by the decay-corrected point-source activity.

2.5. Tomographic sensitivity

In order to compare the D-SPECT sensitivity with that of a conventional scanner including the region-centric component, we performed a series of point-source acquisitions using the scanning pattern from three patient scans with different ROI sizes—large, average and small. D-SPECT scans were performed with a point source placed at various locations in the FOV. The same point source was also scanned on the Infinia with LEHR collimators and a standard cardiac acquisition protocol. The point source was placed in the area where the heart would have been in a patient scan on the Infinia.

The tomographic sensitivity was calculated as the background-subtracted total number of counts divided by the acquisition time and the point-source activity, determined as described in the previous section. For each scanning pattern on the D-SPECT, the system generates a relative sensitivity distribution map for the whole FOV. This map was scaled, based on the point-source measurements, to obtain an absolute sensitivity distribution map which was used for the calculation of mean sensitivities in different regions. Since the sensitivity within the pre-defined ROI was not quite uniform, we report the mean sensitivity both over the pre-defined ROI as well as over a smaller concentric ROI representing a sample of a region of approximately uniform high sensitivity. The maximum sensitivity in the FOV is also reported.

2.6. Count rate performance

The count rate performance was studied for one of the detector arrays (#0) using a planar source filled with 99mTc with scattering media (perspex) in front (∼8 cm) of as well as behind (∼2 cm) the source. The activity was ∼7.5 GBq at the start, and scans were made every hour over a 60 h period. The count rate performance of the system was investigated by plotting the measured count rate as a function of expected count rate, obtained by extrapolation of the count rate at the end of the study with decay correction.

2.7. Intrinsic resolution

The intrinsic detector resolution was estimated from a series of system resolution measurements at different distances. A 57Co rod source was placed vertically at different
positions in the FOV and scanned in the sweep mode. For each source position and each detector, the projection angle at which the central ray was perpendicular to the detector surface was selected. The data were integrated in the vertical direction, and a Gaussian function was fitted to the resulting profile. The FWHM values of the fitted curves were plotted as a function of source-to-detector distance and fitted to the function:

$$R_s = \sqrt{R_i^2 + (kx)^2},$$

where $R_s$ is the system resolution, $R_i$ is the intrinsic resolution, $k$ is a constant, and $x$ is the source-to-detector distance. The data were fitted in a least-squares sense, with $k$ and $R_i$ as free parameters. In order to determine the uncertainty in the estimated parameters, we used a bootstrap approach (Buvat 2002). One hundred bootstrap data sets were generated, based on the fit of the original data, by adding normally distributed random noise with a standard deviation equal to that of the residuals of the original fit. This was only done for the D-SPECT data.

### 2.8. Reconstructed resolution

Since the D-SPECT data are reconstructed with OSEM, which is a nonlinear method, the final resolution is dependent on position and on the activity distribution within the FOV. In order to estimate the reconstructed resolution in a clinical imaging situation, we used a perturbation technique (Stamos et al 1988, Liow and Strother 1993, Du et al 2005) and an anthropomorphic phantom. The phantom consisted of an elliptical perspex cylinder representing a human thorax, which contained lung compartments as well as a cardiac insert (Data Spectrum, Hillsborough, USA). Perturbation studies were performed by doing two separate scans: one scan with a line source inside the cardiac insert and no activity in the rest of the phantom, and one scan without the line source but with an activity distribution corresponding to a normal myocardial study. A composite data set was then generated by adding a scaled version of the line-source data to the myocardial data. The data sets with and without line sources were reconstructed separately and subsequently subtracted, resulting in an image of the line source only. In this way, the FWHM of the line source will be representative of the resolution in the myocardial image. Separate perturbation studies were done on the D-SPECT and on the Infinia.

#### 2.8.1. D-SPECT

A flexible line source (diameter 1 mm, length 20 cm), filled with 10 MBq of $^{99m}$Tc, was placed inside the myocardial compartment. The source was taped to the inner wall in a U-shape bending around the apex. The source was placed at such an angle that it would be confined to a horizontal plane when the phantom was in the upright position, although the heart itself was at an anatomically correct angle. This was done in order to avoid the interpolation associated with image realignment, which would have influenced the measured resolution. Data were acquired with the D-SPECT for 6 min in the sweep mode. The line source was then removed and the myocardial compartment filled with a uniform distribution of $^{99m}$Tc (10 MBq, 110 mL). The phantom background was filled with a solution of 60 MBq of $^{99m}$Tc in ~7 L of water. The ventricular cavity was filled with a solution of concentration similar to that of the background. The phantom was replaced in the same position in the D-SPECT scanner, and a second scan was performed for 30 min.

The line-source data were scaled by a factor of 0.05 and added to the myocardium data. Since the myocardial scan was five times longer, the effective scaling factor for the line-source data was 0.01. The composite data set and the myocardial data were reconstructed separately using two different protocols: (1) without smoothing, and (2) with standard clinical smoothing applied between iterations. No scatter or attenuation corrections were applied. The
Performance evaluation of D-SPECT

reconstructed voxel size was 4.92 mm cube. Horizontal and vertical profiles were generated through the two straight sections of the U-shaped line source. Gaussian functions were fitted to the profiles and the FWHM value determined and averaged over the two line-source sections.

2.8.2. Infinia. Separate scans were performed on the Infinia. This time the line source (containing 10 MBq of $^{99m}$Tc) was aligned with the long axis of the heart. In the second scan, the myocardium was filled with 10 MBq of $^{99m}$Tc, uniformly distributed, but no activity was added to the rest of the phantom. In each scan, data were acquired for 15 min using a standard clinical protocol and LEHR collimators.

A composite data set was generated by adding the line-source data, scaled by a factor of 0.01, to the myocardium data. The three data sets were reconstructed by FBP into a matrix that was tilted in such a way that the line source would be parallel to a coronal plane in the reconstructed 3D volume. This was achieved using realignment parameters determined for an initial reconstruction with a non-tilted image matrix. This procedure was used in order to avoid the interpolation associated with a separate realignment step. No scatter or attenuation corrections were applied. The voxel size was 2.4 mm cube. The reconstruction was done with a ramp filter and subsequently a Butterworth post-filter, $H(\mathbf{f}) = [1 + (\mathbf{f}/f_c)^2]^n^{-1/2}$, was applied. The filter parameters used were average values from the EANM/ESC guidelines for nuclear cardiology (Hesse et al. 2005): $n = 6$, $f_c = 0.42$ cm$^{-1}$. FWHM values determined for the line-source image obtained by the perturbation procedure and for that reconstructed directly from the acquired line-source data were identical.

The data were also reconstructed using an iterative algorithm including resolution recovery: Evolution (GE Healthcare, Haifa, Israel). For this purpose, two new data sets were generated: (1) the line-source data plus the myocardial data, scaled by a factor of 20 and (2) myocardial data only, scaled by a factor of 20. The reason for scaling up the myocardial data rather than scaling down the line-source data was that a 2-byte integer file format was used. The line-source image, obtained by subtraction of the two reconstructed images, was re-oriented using trilinear interpolation. The effect of the interpolation was estimated by applying the same transformation to an image with only one non-zero voxel, fitting a Gaussian function to the result in three different dimensions and calculating the average FWHM. Correction was done by subtracting this value from the measured FWHM value in quadrature. The voxel size was 4.4 mm cube.

2.8.3. Voxel size correction. In order to take into account the effect of the relatively large voxel size on the estimated resolution, we performed 2D computer simulations by generating a Gaussian function on a finely sampled grid, re-sampling the data into grids with 2.4 mm or 4.92 mm pixels and fitting a Gaussian function to the re-sampled data sets. With the 2.4 mm grid the resolution was overestimated by 1% at a true FWHM of 10 mm. With the 4.92 mm grid the resolution was overestimated by 10%, 5% and 2.5% at true FWHM values of 7.5 mm, 10 mm and 15 mm, respectively. We used these figures to correct the bias in measured line-source resolutions.

2.9. Jaszczak phantom

Although the D-SPECT has been designed specifically for cardiac imaging, it is of interest to investigate its general imaging performance characteristics. For this purpose, we chose a phantom which would reflect the resolution of the system across the FOV.

The flangeless Deluxe Jaszczak phantom (Data Spectrum, NC), which contains six sections of cold rods with diameters 12.7, 11.1, 9.5, 7.9, 6.4 and 4.8 mm, was scanned twice.
on the D-SPECT and once on the Infinia. The phantom was filled with $^{99m}$Tc, and the total activity at the start of each scan was: 365 MBq (D-SPECT scan 1), 339 MBq (D-SPECT scan 2) and 313 MBq (Infinia). Data were acquired for 30 min in each scan. The two D-SPECT scans were acquired with the phantom in different orientations relative to the scanner; with the second largest and third largest rod sections closest to the corner of the scanner, respectively. This was done since the image quality is expected to be the best close to the scanner. The D-SPECT data were reconstructed with and without filtering. The Infinia data were reconstructed by FBP with a ramp filter, as well as with Evolution (two iterations, ten subsets). A 3D Hamming post-filter (cut-off frequency $= 1.15 \text{ cm}^{-1}$) was applied to the images. A simple attenuation correction (Chang 1978) was also applied. Transaxial slices from the cold-rod part of the phantom were integrated over a range of $\sim 20 \text{ mm}$ along the $z$-axis, and analysed by visual inspection.

3. Results

3.1. Energy resolution

The energy spectra for the five radionuclides $^{57}$Co, $^{99m}$Tc, $^{123}$I, $^{153}$Gd and $^{201}$Tl are shown in figure 2(a). The two $^{57}$Co peaks (122 and 136 keV) are clearly distinguished as are two
Performance evaluation of D-SPECT

Figure 3. D-SPECT tomographic sensitivity distribution, as specified by the system, for three different ROI sizes: large (a), average (b) and small (c). The larger ellipse is the ROI selected, while the smaller ellipse represents a sample of the region of approximately uniform high sensitivity. The colour scale goes from 0 to the maximum sensitivity for each distribution.

low-energy 201Tl peaks, composed of several photon energies (approximately 70 keV and 80 keV). In the 201Tl spectrum, two more peaks can be seen at higher energies at 135 and 167 keV. There are also peaks at 59 keV and 67 keV present in all spectra—these are due to x-rays from the tungsten collimators. The energy linearity, in terms of peak channel versus photon energy, of the detectors was very good (data not shown). It is noted that the photopeaks have larger low-energy tails than might be expected on conventional single-crystal cameras. These arise due to the properties of pixelated CZT detectors where there are several mechanisms whereby reduced energy is measured (Wagenaar 2004).

The energy resolution (FWHM) as a function of photon energy is shown in figure 2(b) for the D-SPECT as well as for the Prism 3000. For the D-SPECT at 70 keV the resolution was 8.3% and 10.8%, as determined by Gaussian fitting and direct measurement, respectively, and at 140 keV, 5.1% and 5.5%, respectively. Compared with the NaI detectors, there is a ratio in the range 1.4–2.0, depending on the photon energy and estimation method. Figure 2(c) shows histograms of individual D-SPECT pixel resolution values. There was quite a wide range, with a factor of ∼2 between the best and the worst pixels.

3.2. Scatter fraction

The D-SPECT reconstructed scatter fraction was estimated to be 30% for 99mTc with a 20% energy window in a 20 cm cylindrical phantom. For the Infinia the corresponding value was 34%, which is similar to previously reported scatter fractions for NaI-based SPECT systems (see, e.g., Msaki et al (1993)).

3.3. Detector sensitivity

The activity in the point source was 24.5 MBq and the measured count rate was 9743 s$^{-1}$ (background subtracted), giving an absolute sensitivity per head for 99mTc of $S = 398$ s$^{-1}$ MBq$^{-1}$, which can be compared to the values of 122 s$^{-1}$ MBq$^{-1}$ and 72 s$^{-1}$ MBq$^{-1}$ for a standard gamma camera with a 3/8” NaI crystal and a LEGP, and a LEHR collimator, respectively.

3.4. Tomographic sensitivity

Figure 3 shows the relative sensitivity distribution in the FOV of the D-SPECT with the three different scanning patterns used together with the pre-defined and central ROIs. The
Figure 4. D-SPECT count rate performance; measured versus expected count rate, obtained with a decaying $^{99m}$Tc source. The dashed line is the line of identity.

Table 2. D-SPECT sensitivity for $^{99m}$Tc for three different scanning patterns, absolute sensitivity (s$^{-1}$ MBq$^{-1}$) (relative sensitivity compared to the Infinia).

<table>
<thead>
<tr>
<th></th>
<th>Pattern 1</th>
<th>Pattern 2</th>
<th>Pattern 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>ROI</td>
<td>647 (4.6)</td>
<td>747 (5.3)</td>
<td>874 (6.2)</td>
</tr>
<tr>
<td>C-ROI$^a$</td>
<td>729 (5.2)</td>
<td>887 (6.3)</td>
<td>1107 (7.9)</td>
</tr>
<tr>
<td>Max</td>
<td>792 (5.6)</td>
<td>957 (6.8)</td>
<td>1174 (8.3)</td>
</tr>
</tbody>
</table>

$^a$ Central ROI representing a region of approximately uniform sensitivity.

tomographic sensitivity of the Infinia was 141 s$^{-1}$ MBq$^{-1}$. The mean tomographic sensitivities of the D-SPECT for the two ROIs as well as the maximum values are presented in table 2 for the three scanning patterns. Relative sensitivity values compared to the Infinia are also given. The D-SPECT tomographic sensitivity varies significantly depending on the size of the heart and its position in the FOV as well as on the operator who draws the ROI. The maximum sensitivity values range from 792 to 1174 s$^{-1}$ MBq$^{-1}$, and the ROI sensitivities from 647 to 1107 s$^{-1}$ MBq$^{-1}$, corresponding to relative sensitivities of 4.6–7.9 compared to the Infinia.

3.5. Count rate performance

Figure 4 shows the measured count rate versus the expected count rate for one single detector array, indicating no count loss up to a rate of 160 k s$^{-1}$ after which the system becomes saturated with no further increase. The corresponding curve for a conventional scintillation detector based camera would show a more gradual count rate loss (see, e.g., Strand and Larsson (1978)). A 20% loss at 300 k s$^{-1}$ per head was specified for the Infinia (GE Healthcare, Haifa, Israel). The difference in high count rate response is related to the signal processing electronics as well as the pixelation of the D-SPECT detectors. Since the D-SPECT contains nine detector arrays, the maximum system count rate is approximately $9 \times 160$ k s$^{-1} = 1.44 \text{ M s}^{-1}$. 
3.6. Intrinsic resolution

Figure 5 shows the measured system resolution of the two systems as a function of distance from the detector surface together with the fitted functions. Due to the system geometry it was not possible to place the point source closer than $\sim 62$ mm. It can be seen that, for the D-SPECT, the fitted function is actually a straight line, implying an intrinsic resolution of 0. The bootstrap procedure resulted in $R_i = 0.70 \pm 0.77$ mm (mean $\pm$ SD). This means that the actual intrinsic resolution is equal to the detector pixel spacing (2.46 mm).

For the Infinia, the estimated intrinsic resolution was 3.8 mm, which is the same as the value specified by the manufacturer (GE Healthcare, Haifa, Israel), showing that this method can produce reasonable results.

3.7. Reconstructed resolution

D-SPECT images from the perturbation experiment reconstructed with smoothing are shown in figure 6(a), including myocardial images with and without the line source, as well as the subtraction of the two. A horizontal profile through the two sections of the line source is shown in figure 6(b) together with fitted Gaussian functions. Figure 6(c) shows the estimated resolution as a function of the number of reconstruction updates (iterations times subsets). For the standard protocol without smoothing, the estimated horizontal and vertical resolutions (FWHM) were 8.4 mm and 13.8 mm, respectively, and with smoothing, 10.4 mm and 14.6 mm, respectively (after voxel-size correction). Improved resolution can be obtained by using more reconstruction iterations (figure 6(c)).

For the Infinia, the measured FWHM value was $10.5 \pm 1.5$ mm for FBP reconstruction with a ramp filter (mean $\pm$ SD). With the Butterworth post-filter the result was FWHM = 13.7 $\pm$ 1.3 mm. Clinical reconstruction can also be performed using the recently introduced option, Evolution, which utilizes OSEM incorporating a resolution model. In this case the measured FWHM was 10.6 mm, with 12 iterations, 10 subsets and no post-filter. This value is similar to that obtained with the FBP and ramp filter. The convergence was faster than for the D-SPECT (figure 6(c)), as expected with high-resolution collimators.
3.8. Jaszczak phantom

Reconstructed images from the Jaszczak phantom are shown in figure 7(a) for the D-SPECT and 7(b) for the Infinia. In the first D-SPECT scan, the rods in the three first sections (diameters 12.7, 11.1 and 9.5 mm) are clearly visualized both with and without filtering. When the phantom was rotated by \(\sim 60^\circ\), the rods in the next section (diameter 7.9 mm) could also be distinguished, while some of the largest rods became a bit fuzzy. This shows that it is
important to position the patient with the heart in the ‘near-field’ of the scanner. In the Infinia scan, the phantom orientation corresponds approximately to that of the first D-SPECT scan. Also here the rods in the first three sections can be distinguished with both FBP and Evolution. There are some artefacts near the edge of the phantom which are present in the D-SPECT images as well as in the Infinia/Evolution images but not in the Infinia/FBP ones. This would suggest that these artefacts are related to the resolution-recovery algorithms, perhaps due to lack of proper attenuation correction.

4. Discussion

The D-SPECT system utilizes a novel design, which offers opportunity for region-centric acquisition unlike conventional SPECT systems. The system is designed specifically for cardiac SPECT use, and therefore conventional planar measurements are not offered as standard options. The challenge has been to demonstrate not only the intrinsic capabilities of the system but also to show how this translates into clinical operation. National Electrical Manufacturers Association (NEMA) procedures (2007) have been developed as standard means of specifying instrument performance but these are specific to conventional systems, and are not necessarily applicable to systems with an alternative design. Consequently, direct comparison of measurements with those obtained in conventional systems is not necessarily meaningful. We have attempted to perform a set of objective measurements to reflect performance; some of these measurements conform closely with NEMA procedures, others do not. However, these may form the basis for future consideration of performance measures that would be appropriate for non-conventional instrumentation.

The D-SPECT system demonstrates a significant improvement in sensitivity compared to conventional systems, due to the combined wide-angle collimator and region-centric acquisition. There are a number of alternative approaches to achieving sensitivity gain that have been recently suggested; these include use of multiple pinholes (Funk et al 2006, Beekman and van der Have 2007), multiple segment slant hole collimators (Xu et al 2007) and slit-slat collimators as implemented in the CardiArc and MarC systems (Chang et al 2006, Patton et al 2007). We made no attempt to directly compare the relative merits of these alternative systems. Instead, we have attempted to demonstrate the count improvement both in terms of absolute measurement in planar and tomographic modes, which we believe provides evidence of the practical gain in sensitivity. Clearly, reconstructed noise is dependent not only on the acquisition statistics but on reconstruction parameters; as always, there is a balance between reconstructed noise and image contrast (which reflects the reconstructed resolution). In the paper, we limited our assessment to the use of standard reconstruction parameters recommended for clinical use.

Estimation of reconstructed resolution is a non-trivial exercise, especially for systems with iterative reconstruction as an intrinsic component. While intrinsic resolution can be estimated from the measurement of projections, the reconstructed resolution, with collimator modelling included, no longer conforms to standard NEMA protocols for measurement. Measurements, therefore, have little meaning in direct comparison to NEMA specifications for conventional systems. Furthermore, the influence of the system model during reconstruction is object dependent and so the performance must be assessed for a realistic object, rather than a simple point or line source. Arguably, resolution is not directly measurable from reconstructed data; however, we have attempted to measure this via an approach that has previously been used in simulation studies. The perturbation approach attempts to measure an ‘effective’ resolution by imposing a negligibly small amount of line-source data on the projections for a physical heart phantom. This permitted us to verify previous simulation results that were previously
reported (Gambhir et al 2009). We recognize the limitations of this approach and therefore additionally included more subjective data illustrating reconstruction of a Jaszczak phantom. Incidently, the intrinsic resolution was estimated directly from extrinsic measurements, an approach that could easily be applied in conventional systems, removing the need for a somewhat cumbersome direct measurement of intrinsic resolution using a collimated source (which has associated risks of crystal damage).

The specific advantage of the CZT detectors is apparent from the illustrated energy spectra. Energy resolution is far superior to that obtained with NaI (in terms of FWHM), with good energy separation for the multiple radionuclides used. A consequence of this is the possibility of using multiple radionuclides simultaneously. Correction for spillover is still needed due to scatter from the upper energy radionuclide being detected in the energy window selected for the lower-energy radionuclide. In addition, there is a need to correct for the spillover that results from the broad photopeak tails that are present in the pixelated CZT detector (Kacperski et al 2008). Note however that the sensitivity advantage of the system permits a combination of reduced activity and extended acquisition which can reduce the spillover fraction. The sensitivity gain also provides scope for short acquisition dynamic data; in addition, the excellent linear count rate performance ensures few dead-time losses during high activity passage in the early period after tracer administration.

Our results show that, compared to the GE Infinia with LEHR collimators, the D-SPECT has better energy resolution (by a factor of approximately 2), higher sensitivity (by a factor of 5–8 or higher) and similar spatial resolution. However, we believe that, in its current configuration, the D-SPECT may not be operating at an optimal level. The superior energy resolution of the CZT detectors (5% at 140 keV) could be utilized better by using narrower energy windows (currently 20%), which would lead to a reduced scatter fraction (currently only slightly better than a NaI system, 30% versus 34% in a 20 cm diameter phantom) without losing too much sensitivity. Recent theoretical findings regarding reconstruction from limited data sets (Defrise et al 2006) suggest that the fraction of time the detectors spend viewing the pre-selected ROI could perhaps be increased, leading to higher overall sensitivity. The reconstructed resolution with the clinical D-SPECT protocol (approximately 12 mm) is similar to that of the Infinia with FBP reconstruction, but seems to degrade rapidly outside the near-field of the scanner. Currently, the reconstructed voxel size (4.92 mm) is twice the detector pixel size. A reduced voxel size could potentially lead to improved image quality. The addition of some kind of transmission scanning capability, and thereby the possibility of non-uniform attenuation correction, should also be beneficial.

5. Conclusions

We have presented a set of objective measurements performed on a pre-release version of the D-SPECT system. These measurements demonstrate that the system has superior sensitivity to that of conventional dual-headed cameras, while achieving equal or better resolution. The D-SPECT system has excellent energy resolution which should prove useful in dual radionuclide studies. The system also has excellent count rate characteristics which, combined with high sensitivity, should facilitate high-quality dynamic acquisition.

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