Relative Impact of Scatter, Collimator Response, Attenuation, and Finite Spatial Resolution Corrections in Cardiac SPECT

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We determined the relative effect of corrections for scatter, depth-dependent collimator response, attenuation, and finite spatial resolution on various image characteristics in cardiac SPECT.

Methods: Monte Carlo simulations and real acquisition of a 1400Tc cardiac phantom were performed under comparable conditions. Simulated and acquired data were reconstructed using several correction schemes that combined different methods for scatter correction (3 methods), depth-dependent collimator response correction (frequency-distance principle), attenuation correction (nonuniform Chang correction or within an iterative reconstruction algorithm), and finite spatial resolution correction (use of recovery coefficients). Five criteria were considered to assess the effect of the processing schemes: bull’s-eye map (BEM) uniformity, contrast between the left ventricle (LV) wall and the LV cavity, spatial resolution, signal-to-noise ratio (SNR), and percent errors with respect to the known LV wall and liver activities. Results: Similar results were obtained for the simulated and acquired data. Scatter correction significantly improved contrast and absolute quantitation but did not have noticeable effects on BEM uniformity or on spatial resolution and reduced the SNR. Correction for the depth-dependent collimator response improved spatial resolution from 13.3 to 9.5 mm in the LV region, improved absolute quantitation and contrast, but reduced the SNR. Correcting for attenuation was essential for restoring BEM uniformity (78% and 89% without and with attenuation correction, respectively [ideal value being 100%]) and accurate absolute activity quantitation (errors in estimated LV wall and liver activity decreased from 90% without attenuation correction to ~20% with attenuation correction only). Although accurate absolute activity quantitation was achieved in the liver using scatter and attenuation corrections only, correction for finite spatial resolution was needed to estimate LV wall activity within 10%. Conclusion: The respective effects of corrections for scatter, depth-dependent collimator response, attenuation, and finite spatial resolution on different image features in cardiac SPECT were quantified for a specific acquisition configuration. These results give indications regarding the improvements to be expected when using a specific processing scheme involving some or all corrections.

Key Words: cardiac SPECT; quantitation; Monte Carlo simulations; scatter; depth-dependent collimator response; attenuation; finite spatial resolution

J Nucl Med 2000; 41:1400-1408

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catter, depth-dependent collimator response, attenuation, and finite spatial resolution are major factors affecting cardiac SPECT and have received substantial attention in the past years. Many correction methods have been proposed to compensate for scatter (1), depth-dependent collimator response (2,3), and attenuation (4), whereas few studies have been devoted to the issue of finite spatial resolution (FSR) in SPECT (5,6). Although the evaluation of a single correction or of a combination of corrections in cardiac SPECT has been reported—e.g., scatter (7,8), attenuation (9,10), attenuation and variable collimator response (11), attenuation and scatter (12), attenuation, scatter, and variable collimator response (13)—there is a lack of data regarding the respective contribution of the different corrections to the final image characteristics.

Using Monte Carlo simulations and real acquisitions of a cardiac phantom, we previously determined how scatter, depth-dependent collimator response, attenuation, and FSR impacted the image features in cardiac SPECT (14). This allowed us to predict which corrections should be essential for the recovery of uniformity, contrast, spatial resolution, signal-to-noise ratio (SNR), and accurate quantitation. In this study, we have considered available correction methods for scatter, depth-dependent collimator, attenuation, and FSR and incorporated them into various processing schemes on the same simulated and acquired cardiac data to determine how current corrections compare with ideal corrections and to assess their respective impact on the image characteristics.

MATERIALS AND METHODS

Cardiac Phantom

A 24 × 32 cm elliptic cardiac phantom (Data Spectrum, Chapel Hill, NC) was scanned using an MRI scanner, and the resulting MR images were segmented into a 10-mm-thick left ventricle (LV)
wall, 2 lung compartments, a liver, a Teflon spine, and a soft-tissue compartment as described (14). Narrow-beam, energy-dependent attenuation values were assigned to the different compartments (15). These numeric data were used to perform Monte Carlo simulations while acquisitions of the same phantom were obtained under similar conditions so that results from simulations and acquisitions could be compared.

**Acquisitions**

The LV wall and the liver compartment of the cardiac phantom were filled with 79 kBq/mL and 47 kBq/mL 99mTc, whereas the soft-tissue compartment contained only water. A SPECT acquisition (2.3 million counts in the 20% energy window) was performed using a dual-head DST camera (Sophia Medical Vision International, Buc, France) equipped with a low-energy, high-resolution (LEHR) collimator. One hundred twenty-eight projections were acquired over 360° in 3 energy windows (93-122, 123.5-128.5, and 126-154 keV) using a 20.5-cm circular orbit. The pixel size was 3.8 × 5 mm in the projections. The phantom was left at the same position, and a transmission scan was obtained 24 h later when the residual 99mTc activity could be considered as negligible. The transmission acquisition was performed using 2 collimated 153Gd (100 keV) scanning line sources (3.7 GBq each) facing the 2 collimators at 90°. Sixty-four transmission projections were acquired over a 180° circular orbit in 64 min in the 90- to 110-keV energy window. The transmission projections were reconstructed using filtered backprojection (FBP) (Hann filter; v = 0.5 cycle/pixel). The resulting 153Gd attenuation coefficients (μ153Gd) were linearly scaled to the 99mTc attenuation coefficients (μ99mTc) using μ153Gd = 0.9 × μ99mTc.

**Monte Carlo Simulations**

Monte Carlo simulations were performed using the SimSET software (PHG 2.04b) (16), to which modeling of coherent scatter was added (17). Scattered photons were tracked until the ninth order. Only the geometric component of the LEHR collimator response was modeled. The intrinsic energy response was modeled using Gaussian functions with SDs dependent on energy (energy resolution set to 9.5% at 140 keV). Two simulations were performed. In a first simulation (air simulation), activity was set only in the LV wall and no attenuating medium was simulated, as if the photons propagated in air. In a second simulation, activity was added to the LV wall and in the liver, and attenuation of the different tissues was accounted for, to mimic the real SPECT acquisition. The same acquisition geometry and parameters as those of the real acquisition were used. Forty-one million photons were detected in the air simulation, and 8 million photons were detected in the 20% energy window when simulating the whole phantom. The Monte Carlo events were sorted according to their energy, detected position (x, y, z), and type (primary or scattered). The resulting data were arranged into 128 projections, 128 × 44. Each projection included 28 spectral images, each 3.5-keV wide, from 63 to 161 keV. No transmission acquisition was simulated.

**Correction Methods**

The correction methods are described in the order they are applied in the different processing schemes (Fig. 1).

**Scatter Correction Methods.** In addition to the 20% energy window (120) images (126-154 keV), 3 scatter corrections were investigated: the Jaszczak subtraction (JAS), the triple-energy-window (TEW) method, and a spectral factor analysis (SFA). All scatter corrections were performed on the projections before tomographic reconstruction.

For each projection, the JAS image was obtained by subtracting the 91- to 122.5-keV Compton image, weighted by k = 0.5, from the 120 image (18).

The TEW method (19) estimated the scatter-free image I_{TEW} using the number of photons detected in 2 3.5-keV images, I_1(122.5-126 keV) and I_2(150.5-154 keV), by:

\[ I_{TEW} = I_2 - (I_1 + I_2) \times \frac{w}{2w'}, \]

where w was the energy width of the photopeak window (28 keV) and w' was 3.5 keV. No noise filtering was performed on images I_1 and I_2 before subtraction from the 20% window.

For each projection, SFA used 28 3.5-keV-wide spectral images (63-161 keV) to estimate the scatter-free projections using a linear model (20,21). This model assumes that the energy spectrum of the photons detected in each pixel is the weighted sum of a photopeak and 2 Compton spectra. The photopeak is estimated using a target apex-seeking procedure (22), whereas the scatter spectra are estimated using an iterative approach, with constraints pertaining to the theoretic non-negativity of spectra and associated images. Only

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**FIGURE 1.** Processing schemes applied to simulated and acquired data.
1 SFA was performed dealing with all projections simultaneously and resulting in a set of 128 scatter-free projections. SFA was used for only the simulated data and not for the acquired data because no SPECT multispectral acquisition device was available on our camera. In addition, for the simulated data, an "ideal" scatter correction was obtained by considering only the primary photons.

**Depth-Dependent Collimator Response Correction.** The projections were corrected for depth-dependent collimator response using the frequency–distance principle (FDP) (2). The inverse deconvolution filter was implemented using the measured full width at half maximum (FWHM) of the collimator point spread function (PSF) used in the acquisition and simulation. The cutoff frequency of the inverse filter was controlled by a dimensionless parameter, , set to . For the simulated data, images corresponding to an ideal correction for the depth-dependent collimator response were obtained by blurring the original slices of activity distribution using a Gaussian kernel with an FWHM equal to that of the collimator PSF at the face of the detector (FWHM = 6.6 mm) and adding Poisson noise to the blurred images.

**Attenuation Correction and Reconstruction.** When no attenuation correction was used, reconstruction was performed using FBP (Hann filter; , cycle/pixel). Two nonuniform attenuation corrections were investigated: a nonuniform, second-order Chang correction (CH) combined with FBP (Hann filter; , cycle/pixel) (23,24) and a correction within an iterative reconstruction algorithm. The iterative correction was performed using 20 iterations of a least-square minimal residual algorithm (MR) (25), with modeling of nonuniform attenuation only in the projector (not in the backprojector). Note that performing 1 iteration of MR without attenuation correction is equivalent to using FBP with a Hann filter.

For the simulated data, the nonuniform attenuation correction was performed using the exact attenuation map used for the simulation. For the acquired data, the measured attenuation map was used. Ideal correction for attenuation was obtained using the Monte Carlo simulation of the LV compartment in air.

**FSR Correction.** Even when depth-dependent collimator response correction was performed, spatial resolution in the reconstructed images was limited, yielding an underestimation of activity in small structures. This effect is called the FSR effect in this article (sometimes also called partial-volume effect in nuclear medicine). FSR was compensated using a simple convolution model (26). A 1-dimensional, 10-mm-wide, square wave was convolved with a 1-dimensional Gaussian of FWHM varying from 6 to 14 mm to predict the underestimation associated with the FSR in a 10-mm-thick LV wall for spatial resolutions (characterized by the FWHM of the PSF) ranging from 6 to 14 mm. For each spatial resolution, the ratio max/s max between the observed maximum of the square wave after convolution (max) and its true maximum (max) was used to deduce the underestimation associated with the FSR in the LV wall, and the recovery coefficient was defined as max/max. The activity measured in the LV wall was multiplied by this recovery coefficient for FSR compensation.

**Data Analysis**

**Evaluation Strategy.** First, the roles of scatter, depth-dependent collimator response, and attenuation corrections were assessed separately by considering the simulated data. To evaluate the scatter correction methods, JAS, TEW, and SFA corrected projections were compared with the simulated projections of primary photons. The performance of FDP collimator response correction alone was studied by comparing the reconstructed slices of the air-simulated projections with the blurred version of the simulated activity distributions obtained as explained above. To evaluate attenuation correction alone, CH- and MR-reconstructed volumes of the primary photons from the simulation with attenuating media were compared with the FBP-reconstructed volume of the projections obtained from the air simulation. Indeed, these data were not affected by scatter and were affected identically by collimator response and FSR.

Next, we studied the relative impact of each correction on acquired and simulated data by performing incremental correction schemes using the correction methods that yielded the best results in the previous step. Five correction schemes were studied (Fig. 1): attenuation correction alone, scatter + attenuation corrections, collimator response + attenuation corrections, scatter + collimator response + attenuation corrections, and scatter + collimator response + attenuation + FSR corrections.

**Evaluation Criteria.** The effect of the different corrections was characterized using the same 5 criteria as those reported previously (14): uniformity of the bull’s-eye map (BEM), contrast between a cold and a hot region, spatial resolution, SNR, and percent error between estimated and true activity in specific regions.

- Uniformity of BEM. The reconstructed slices were reoriented into short-axis slices, and the BEMs were derived. Each BEM was divided into 9 myocardial regions (14), and the mean activity in each region was calculated. The maximum of these 9 values was used to normalize all 9 values using . The mean of the 9 normalized values defined a uniformity index, which should ideally be 100%, because activity was uniform in the LV wall.

- Contrast. Contrast was calculated between 2 3-dimensional volumes of interest (VOIs) drawn inside the LV wall and cavity. The mean counts in the LV wall and cavity were divided, and the contrast was calculated as . Because there was no activity in the LV cavity, the ideal contrast should be 100%.

- Spatial resolution. To assess spatial resolution, each short-axis slice was divided into 8 sectors, and the radial count profile corresponding to each sector was calculated. Knowing the theoretic thickness and uptake of the wall, the FWHM of the Gaussian, which minimized the mean square error between the observed count profile and the theoretic count profile convolved with the Gaussian, was deduced.

- SNR. A VOI (~1000 voxels) was drawn inside the liver (liver volume was ~7000 voxels in simulated and acquired data) in which the activity was uniform. It was verified using a Kolmogorov-Smirnov test (28) that the pixel values in this VOI were, in all cases, Gaussian distributed, which allowed us to compute an SNR as the mean over the SD of the VOI voxel values. Note that this definition of SNR compares noise level to mean signal value (i.e., the inverse of relative pixel noise) and is not equivalent to the contrast-to-noise ratio.

- Percent error in activity. Using the simulated projections of primary photons, a projection region of interest (ROI) (including pixels from several projections) containing only LV wall activity and a projection ROI containing only liver activity
were drawn. The numbers of counts measured in these ROIs in the different projections (I20, scatter corrected, and primary) were compared. For the simulated and acquired data, the mean activity in each of the 9 regions dividing the BEM was calculated. The average percent errors with respect to the real activity (activity set in the phantom or simulated activity map) over the 9 regions were calculated. The percent error in estimated liver activity was measured in an ~1038-voxel VOI drawn inside the liver. For the acquisition, a cross-calibration factor was measured (8 kcts/pixel/MBq) to convert the number of counts measured in the images into an estimated activity.

RESULTS AND DISCUSSION

First, the impact of the corrections for the simulated data is assessed in terms of the 5 evaluation criteria. Next, the performance of the different correction methods is compared. Finally, the results obtained for simulated and acquired data are compared.

Projections

Because the scatter corrections affected the projections, their accuracy was first characterized by comparing the true number of primary photons in the LV wall and liver projection ROIs with the numbers of photons measured in the projection ROIs of the scatter-corrected images. In the I20 images, the numbers of primary photons in the LV wall and in the liver projection ROIs were overestimated by 46% ± 11% and 47% ± 9%, respectively. Scatter correction significantly reduced this error (Student t test; P < 0.05): SFA and JAS overestimated the number of primary photons by 4% ± 2% and 6% ± 3%, respectively, in the LV wall projection ROI and by 3% ± 1% and 5% ± 2%, respectively, in the liver projection ROI. TEW underestimated this number by 5% ± 3% and 6% ± 2% in the LV wall and liver projection ROIs, respectively. The histogram of the percent errors in the LV wall projection ROI (Fig. 2) showed that SFA yielded the highest number of pixels with an error close to zero: the error was <10% in 56% of the pixels, whereas it was <10% in 0%, 37%, and 39% of the pixels for the I20, JAS, and TEW images, respectively. In the liver projection ROI, the trend was the same (Fig. 3): SFA yielded the highest number of pixels with an error of <10% (66%) followed by TEW (51%), JAS (40%), and I20 (0%).

Reconstructed Volume

Uniformity of LV Activity Distribution. In this work, BEM uniformity characterized relative quantitation between equally hot regions. Attenuation correction was the major correction affecting BEM uniformity: uniformity was 78% with I20 + FBP compared with 88% and 89% with I20 + CH and I20 + MR, respectively (P < 0.05) (Table 1). Scatter correction alone did not improve uniformity (78% for I20 + FBP versus 79% for JAS + FBP or primaries + FBP [not significant]). This results because the LV and the liver were 9 cm apart in our phantom, and therefore only little scatter from the liver activity affected the LV wall region. Similarly,

**FIGURE 2.** Histogram of error with respect to primary distribution in LV projections with I20, SFA, JAS, and TEW.

**FIGURE 3.** Histogram of error with respect to primary distribution in liver projections with I20, SFA, JAS, and TEW.
TABLE 1
Evaluation of Attenuation (A), Scatter (S), Collimator Response (C), and FSR Corrections

<table>
<thead>
<tr>
<th>Corrections</th>
<th>Available or ideal correction method</th>
<th>BEM uniformity (%)</th>
<th>Contrast resolution (mm)</th>
<th>SNR</th>
<th>In LV (ideal = 100%)</th>
<th>In liver (ideal = 0%)</th>
<th>In LV after FSR compensation (ideal = 0%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>None</td>
<td>I20 + FBP</td>
<td>79.8 ± 13.8</td>
<td>0.60</td>
<td>13.3 ± 1.1</td>
<td>4.7</td>
<td>-91.8 ± 1.5</td>
<td>-85.7 ± 2.5</td>
</tr>
<tr>
<td>A</td>
<td>I20 + CH</td>
<td>87.8 ± 9.5</td>
<td>0.61</td>
<td>13.2 ± 0.9</td>
<td>4.8</td>
<td>-14.5 ± 10</td>
<td>23.8 ± 3.2</td>
</tr>
<tr>
<td>A</td>
<td>I20 + MR</td>
<td>89.0 ± 7.1</td>
<td>0.63</td>
<td>13.1 ± 0.9</td>
<td>5.0</td>
<td>-21.7 ± 4.8</td>
<td>15.7 ± 2.4</td>
</tr>
<tr>
<td>S</td>
<td>Primaries + FBP</td>
<td>79.3 ± 14.3</td>
<td>0.73</td>
<td>13.1 ± 1.0</td>
<td>4.3</td>
<td>-93.2 ± 2.0</td>
<td>-89.1 ± 2.2</td>
</tr>
<tr>
<td>S</td>
<td>JAS + FBP</td>
<td>91.5 ± 15.4</td>
<td>0.69</td>
<td>13.0 ± 1.7</td>
<td>4.3</td>
<td>-92.8 ± 2.5</td>
<td>-87.8 ± 3.6</td>
</tr>
<tr>
<td>S</td>
<td>TEW + FBP</td>
<td>79.8 ± 14.5</td>
<td>0.68</td>
<td>13.1 ± 1.5</td>
<td>3.9</td>
<td>-93.6 ± 2.3</td>
<td>-89.6 ± 3.4</td>
</tr>
<tr>
<td>S</td>
<td>SFA + FBP</td>
<td>80.4 ± 12.6</td>
<td>0.69</td>
<td>13.1 ± 1.4</td>
<td>4.3</td>
<td>-92.0 ± 2.0</td>
<td>-89.1 ± 2.2</td>
</tr>
<tr>
<td>S + A</td>
<td>Air + FBP</td>
<td>95.6 ± 3.0</td>
<td>0.90</td>
<td>12.6 ± 0.9</td>
<td>7.9</td>
<td>-34.4 ± 1.8</td>
<td>-50.0 ± 2.1</td>
</tr>
<tr>
<td>S + A</td>
<td>Primaries + CH</td>
<td>92.6 ± 5.7</td>
<td>0.73</td>
<td>12.6 ± 1.0</td>
<td>4.4</td>
<td>-28.4 ± 3.7</td>
<td>-5.4 ± 2.9</td>
</tr>
<tr>
<td>S + A</td>
<td>Primaries + MR</td>
<td>94.6 ± 4.3</td>
<td>0.77</td>
<td>12.5 ± 1.0</td>
<td>4.8</td>
<td>-34.4 ± 4.5</td>
<td>-6.3 ± 2.7</td>
</tr>
<tr>
<td>S + A</td>
<td>JAS + CH</td>
<td>88.8 ± 9.8</td>
<td>0.70</td>
<td>12.5 ± 0.9</td>
<td>4.2</td>
<td>-25.1 ± 14.5</td>
<td>3.6 ± 3.9</td>
</tr>
<tr>
<td>S + A</td>
<td>JAS + MR</td>
<td>91.3 ± 6.6</td>
<td>0.73</td>
<td>12.4 ± 0.8</td>
<td>4.2</td>
<td>-31.4 ± 5.3</td>
<td>-3.4 ± 2.9</td>
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<tr>
<td>S + A</td>
<td>TEW + CH</td>
<td>94.4 ± 4.3</td>
<td>0.69</td>
<td>12.6 ± 1.2</td>
<td>3.3</td>
<td>-29.2 ± 3.1</td>
<td>-6.4 ± 3.1</td>
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<tr>
<td>S + A</td>
<td>TEW + MR</td>
<td>93.8 ± 5.3</td>
<td>0.72</td>
<td>12.5 ± 1.1</td>
<td>3.2</td>
<td>-35.5 ± 3.7</td>
<td>-8.2 ± 3.1</td>
</tr>
<tr>
<td>S + A</td>
<td>SFA + CH</td>
<td>93.9 ± 4.6</td>
<td>0.68</td>
<td>12.5 ± 1.7</td>
<td>4.1</td>
<td>-26.9 ± 3.6</td>
<td>-2.0 ± 2.9</td>
</tr>
<tr>
<td>S + A</td>
<td>SFA + MR</td>
<td>92.8 ± 5.3</td>
<td>0.70</td>
<td>12.5 ± 0.9</td>
<td>4.6</td>
<td>-33.2 ± 4.6</td>
<td>-3.6 ± 2.7</td>
</tr>
<tr>
<td>C + A</td>
<td>I20 + FDP + CH</td>
<td>85.8 ± 10.3</td>
<td>0.75</td>
<td>9.6 ± 1.4</td>
<td>3.1</td>
<td>22.2 ± 13.3</td>
<td>26.4 ± 3.7</td>
</tr>
<tr>
<td>C + A</td>
<td>I20 + FDP + MR</td>
<td>90.2 ± 3.0</td>
<td>0.78</td>
<td>9.5 ± 1.3</td>
<td>3.2</td>
<td>-6.6 ± 2.1</td>
<td>18.3 ± 4.1</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>94.4 ± 3.8</td>
<td>0.94</td>
<td>6.6 ± 0.1</td>
<td>14.4</td>
<td>-8.2 ± 3.8</td>
<td>-1.7 ± 1.9</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>95.0 ± 3.3</td>
<td>0.91</td>
<td>9.4 ± 1.2</td>
<td>2.7</td>
<td>-14.4 ± 2.9</td>
<td>-1.9 ± 3.7</td>
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<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>95.7 ± 3.1</td>
<td>0.92</td>
<td>9.4 ± 1.1</td>
<td>2.7</td>
<td>-21.6 ± 2.7</td>
<td>-2.4 ± 2.9</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>91.9 ± 10.8</td>
<td>0.85</td>
<td>9.3 ± 1.5</td>
<td>2.3</td>
<td>-13.3 ± 12.1</td>
<td>5.6 ± 3.2</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>94.6 ± 4.1</td>
<td>0.89</td>
<td>9.2 ± 1.4</td>
<td>2.6</td>
<td>-18.1 ± 3.8</td>
<td>-2.2 ± 2.7</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>95.9 ± 3.6</td>
<td>0.87</td>
<td>9.6 ± 1.5</td>
<td>2.1</td>
<td>-16.2 ± 7.2</td>
<td>3.2 ± 4.1</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>96.2 ± 2.9</td>
<td>0.88</td>
<td>9.5 ± 1.4</td>
<td>2.2</td>
<td>-22.3 ± 2.7</td>
<td>-4.0 ± 3.7</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>94.8 ± 3.1</td>
<td>0.85</td>
<td>9.2 ± 1.4</td>
<td>2.6</td>
<td>-13.8 ± 2.8</td>
<td>1.8 ± 3.2</td>
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<td>S + C + A</td>
<td>Blurred activity map</td>
<td>96.4 ± 2.8</td>
<td>0.86</td>
<td>9.3 ± 1.3</td>
<td>2.7</td>
<td>-21.3 ± 2.6</td>
<td>-1.2 ± 2.1</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>95.3 ± 3.2</td>
<td>0.94</td>
<td>9.0 ± 1.1</td>
<td>5.7</td>
<td>-19.0 ± 2.4</td>
<td>-2.73 ± 2.2</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>100 ± 1</td>
<td>6.6</td>
<td>15 ± 0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>S + C + A</td>
<td>Blurred activity map</td>
<td>100 ± 1</td>
<td>6.6</td>
<td>15 ± 0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>

Ideal values are in bold type.

(14) and with the results of Luo et al. (29) for the MCAT phantom, in which BEM uniformity was 82% in the presence of scatter and attenuation and 98% in air.

Combining collimator response and attenuation corrections did not significantly affect BEM uniformity (uniformity = 89% with I20 + FDP + MR and 90% with I20 + MR). Combining all 3 corrections led to BEM uniformity between 91.9% and 96.4%.

In summary, the key correction for restoring uniform BEM activity was attenuation correction, whereas scatter correction yielded a marginal improvement of uniformity.

Contrast. In our configuration, contrast characterizes relative quantitation between a cold and a hot region. Scatter correction alone significantly improved contrast (0.68 or 0.69 with JAS, TEW, or SFA + FBP compared with 0.60 with I20 + FBP) (Table 1). Collimator response correction also improved contrast in the absence of scatter and attenuation from 0.90 (air + FBP) to 0.94 (air + FDP + FBP). Attenuation correction alone did not significantly improve contrast because differential attenuation was negligible between the LV wall and cavity.

Combining scatter and attenuation corrections slightly improved contrast compared with scatter correction alone, but contrast remained significantly lower than that measured in air + FBP (0.90). Therefore, combined scatter and attenuation corrections led to poorer performance than what would be expected with ideal corrections. For data affected by attenuation, performing collimator response correction in addition to attenuation correction significantly improved contrast (0.75 for I20 + FDP + CH compared with 0.61 for I20 + CH). Indeed, because the cold and the hot regions considered when computing the contrast were close together (~1 cm), enhancing spatial resolution using the FDP increased counts inside the LV wall, reduced counts around the wall, and therefore increased contrast between the LV wall and cavity. The highest contrast values (>0.85) were
obtained when combining scatter, collimator response, and attenuation corrections.

Results obtained with other cardiac and elliptic phantoms and different iterative reconstruction methods confirm the small impact of attenuation correction alone on contrast enhancement, the contrast improvement observed when both scatter and attenuation corrections are performed, and the substantial contrast enhancement observed when collimator response correction is combined with scatter and attenuation corrections (30–32).

Spatial Resolution. Spatial resolution was characterized in the LV region. Neither scatter correction alone nor attenuation correction alone substantially improved spatial resolution in the reconstructed slices (improvement < 0.3 mm; Table 1). On the other hand, correcting for both scatter and attenuation improved spatial resolution slightly but systematically (between 0.7 and 0.9 mm). These results are consistent with the small impact of scatter and attenuation corrections predicted with ideal correction methods (14). The major improvement of spatial resolution was obtained by collimator response correction: combining attenuation and FDP corrections improved spatial resolution by almost 4 mm compared with performing attenuation correction alone (13.1 mm with I20 + MR versus 9.5 mm with I20 + FDP + MR; P < 0.05). This is consistent with the 6-mm improvement (from 15 to 9 mm) reported by Formiconi et al. (33) when including compensation for the variable collimator response in an iterative reconstruction algorithm. Nevertheless, the resulting spatial resolution remained far from the resolution expected at the face of the collimator (6.6 mm). Combining scatter correction to attenuation and FDP corrections did not significantly improve spatial resolution, although there was a trend toward spatial resolution improvement.

These findings suggest that whenever spatial resolution is an important issue (e.g., for the detection of small lesions), performing a collimator response correction can significantly improve spatial resolution.

SNR. Our SNR index characterized primarily high-frequency noise but was by no means intended to characterize lesion detectability. Without any correction (I20 + FBP), the SNR in the reconstructed slices was 4.7 (Table 1). Scatter correction reduced the SNR because all scatter correction methods we used removed counts from the 20% energy window (SNR was 4.3 in JAS or SFA + FBP and 3.9 in TEW + FBP). The SNR was also smaller in the reconstructed slices of the primary projections than in the I20 reconstructed projections (4.3 in primaries + FBP). This increase of relative noise after scatter subtraction has been reported (31) and should be kept in mind when noise is an important parameter. Attenuation correction did not greatly improve the SNR, which remained ≤5, whereas it was 7.9 in air + FBP. Restoring counts is thus not sufficient to restore the high SNR. FDP collimator response correction also reduced the SNR to 3.1 (I20 + FDP + CH or MR) because the FDP operates as an inverse filter, hence increases noise.

The approach of depth-dependent collimator response correction by modeling the collimator spread function in the projector or backprojector operators used in iterative reconstruction could offer a valuable alternative for improving spatial resolution without decreasing the SNR as much (31).

Absolute Activity Quantitation. The corrections required for accurate activity quantitation strongly depended on the size of the considered organ. In the liver, which can be considered as a large organ, the effects of variable collimator response and FSR were negligible, and the major phenomena affecting quantitation were scatter and attenuation. Without any correction, the liver activity was underestimated by 86% with I20 + FBP (Table 1). Correcting for scatter alone did not reduce this error (error > 87%). Correcting for attenuation alone reduced significantly the error (P < 0.05), yielding an activity overestimation of 15%–24%. Attenuation correction had to be combined with scatter correction to achieve a reliable activity estimate in the liver (errors < 8.2%, close to that observed for the air simulation).

For small structures (such as the 10-mm-thick LV wall), activity was also underestimated without any correction (91.8% underestimation with I20 + FBP). Similar to what was observed for the liver, correcting for scatter alone did not improve accuracy (underestimation > 92%). Correcting for attenuation alone reduced the bias (underestimation between 21.7% and 14.5%). Unlike what was observed for large structures, combining scatter and attenuation corrections did not yield accurate quantitation because the underestimation remained between 25% and 36%. These results are consistent with the 28% underestimation measured by Galt et al. (34) in a cardiac phantom for a ratio of wall thickness to system FWHM equal to 1. The residual underestimation after scatter and attenuation corrections is associated with the FSR of the imaging system. Adding FSR compensation to scatter and attenuation corrections gave errors of 2.9%–16.2% depending on the corrections that were used (Table 1). For our configuration, combining collimator response and attenuation corrections yielded an artificial quantitative accuracy for the LV wall activity estimate (errors of −6.6% or +2.2%) because the underestimation of activity associated with FSR compensated almost exactly the activity overestimation caused by scatter. Combining corrections for FSR, collimator response, and attenuation did not yield accurate activity quantitation (error > 27%) because of uncompensated scatter. Combining the FDP with attenuation and scatter corrections reduced the error compared with correcting for scatter and attenuation only because FDP correction improved spatial resolution, hence reducing the FSR effect: error varied between −13% with JAS + FDP + CH and −22% with TEW + FDP + MR. Our results are consistent with those of Gilland et al. (35), who found activity underestimations ranging between 15% and 25% in 3.4- to 21.5-mL spheres after scatter, attenuation, and collimator response corrections. Estimation of LV wall
activity with errors < 10% therefore required correcting for scatter, collimator response, attenuation, and FSR.

**Comparison of Correction Methods**

Comparing results obtained for the different correction methods (Table 1) allowed us to assess the differences in performance seen when correcting the same effects using different approaches and to compare the performance of actual corrections with results predicted for ideal corrections (14). No significant difference was observed between the 3 scatter correction methods in terms of BEM uniformity, contrast between the LV wall and cavity, spatial resolution, or activity quantitation. However, JAS and SFA systematically yielded higher SNRs than did TEW because the 3.5-keV-wide spectral images used by TEW were very noisy and were not smoothed before being subtracted from the photopake image. Smoothing these images, as suggested by Ichihara et al. (36), therefore appears to be necessary to avoid a high level of noise in the scatter-corrected images. Results obtained with the scatter corrections were close to the results obtained using primary photons, meaning the correction methods were reliable enough for this imaging configuration.

The 2 attenuation corrections led to close results in terms of BEM uniformity, contrast, and spatial resolution. The SNR was systematically higher in the MR reconstructed images than with CH. In the absence of scatter, activity quantitation errors were also smaller with primaries + MR than with primaries + CH. In addition, it has been reported that, unlike MR, CH correction was not reliable for data acquired over 180° (37). Therefore, for attenuation correction, MR should certainly be preferred to CH.

The choice of the reconstruction method had a larger impact when combining attenuation correction with scatter or collimator response corrections (or both) than when considering a single correction. Considering scatter + attenuation corrections, uniformity varied between 89% (JAS + CH) and 94% (TEW + CH), contrast varied between 0.68 (SFA + CH) and 0.73 (JAS + MR), SNR varied between 3.2 (TEW + MR) and 4.6 (SFA + MR), and error in estimated liver activity varied between 2% (SFA + CH) and 8% (TEW + MR). When adding FSR compensation (R), the errors in the LV wall activity estimate depended even more on the correction methods that were involved: for instance, LV wall activity was underestimated by 1.2% with SFA + FDP + MR + R, whereas it was underestimated by 9.5% with JAS + FDP + CH + R. In summary, the greater the number of corrections involved, the more critical the choice of each correction method because the imperfections inherent to each correction are amplified differently by the other corrections involved in the whole processing scheme.

The method we used to compensate for FSR gave satisfactory results in our experiments because the LV wall had a constant known thickness. However, the correction as described in this article would not be applicable to real patients because the regional thickness of the wall is usually unknown. Although attenuation and scatter corrections are currently in the stage of clinical evaluation, FSR correction is not yet available but should certainly now be considered to achieve accurate LV activity quantitation (34).

**Comparison of Results Based on Simulated and Acquired Data**

Using the acquired data, scatter correction was performed using JAS, depth-dependent collimator response was corrected using the FDP, attenuation was corrected using MR, and FSR was corrected using recovery coefficients. Numbers of counts were converted into activity values using the measured cross-calibration factor. The values of the evaluation criteria calculated in the images reconstructed from the acquired data are shown in Table 2 (roman type) together with the corresponding values obtained with the simulated data (italic type); good overall agreement is evident. For both acquired and simulated data, attenuation correction was

<table>
<thead>
<tr>
<th>Combined corrections</th>
<th>BEM uniformity (%)</th>
<th>Contrast</th>
<th>Spatial resolution (mm)</th>
<th>SNR</th>
<th>Quantitation error (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>In LV</td>
</tr>
<tr>
<td>None (120 + FBP)</td>
<td>75 ± 13</td>
<td>0.66</td>
<td>13.5 ± 1.2</td>
<td>4.9</td>
<td>-93 ± 3</td>
</tr>
<tr>
<td>Attenuation (120 + MR)</td>
<td>78 ± 14</td>
<td>0.60</td>
<td>13.3 ± 1.1</td>
<td>4.7</td>
<td>-92 ± 2</td>
</tr>
<tr>
<td>Scatter + attenuation</td>
<td>88 ± 8</td>
<td>0.68</td>
<td>13.4 ± 1.5</td>
<td>5.2</td>
<td>-24 ± 5</td>
</tr>
<tr>
<td>(JAS + MR)</td>
<td>89 ± 7</td>
<td>0.63</td>
<td>13.1 ± 0.9</td>
<td>5.0</td>
<td>-22 ± 5</td>
</tr>
<tr>
<td>Scatter + attenuation</td>
<td>88 ± 8</td>
<td>0.77</td>
<td>12.8 ± 2.0</td>
<td>4.4</td>
<td>-28 ± 3</td>
</tr>
<tr>
<td>(JAS + MR)</td>
<td>91 ± 7</td>
<td>0.73</td>
<td>12.4 ± 0.9</td>
<td>4.2</td>
<td>-31 ± 5</td>
</tr>
<tr>
<td>Scatter + attenuation</td>
<td>86 ± 8</td>
<td>0.80</td>
<td>10.2 ± 1.7</td>
<td>4.0</td>
<td>-8 ± 4</td>
</tr>
<tr>
<td>(120 + FDP + MR)</td>
<td>90 ± 3</td>
<td>0.78</td>
<td>9.5 ± 1.3</td>
<td>3.2</td>
<td>-7 ± 2</td>
</tr>
<tr>
<td>Scatter + PSF + attenuation</td>
<td>89 ± 8</td>
<td>0.90</td>
<td>9.6 ± 1.8</td>
<td>3.5</td>
<td>-19 ± 6</td>
</tr>
<tr>
<td>(JAS + FDP + MR)</td>
<td>95 ± 4</td>
<td>0.89</td>
<td>9.2 ± 1.4</td>
<td>2.8</td>
<td>-18 ± 4</td>
</tr>
<tr>
<td>Scatter + PSF + attenuation + FSR</td>
<td>89 ± 8</td>
<td>0.90</td>
<td>9.6 ± 1.8</td>
<td>3.5</td>
<td>-4 ± 3</td>
</tr>
<tr>
<td>(JAS + FDP + MR + R)</td>
<td>95 ± 4</td>
<td>0.89</td>
<td>9.2 ± 1.4</td>
<td>2.8</td>
<td>-2 ± 3</td>
</tr>
</tbody>
</table>

Acquired data are in roman type; simulated data are in italic type.

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the key correction for improving the BEM uniformity, by 13% and 11% in the acquired and simulated data, respectively (comparing I20 + FBP and I20 + MR). Contrast was improved primarily by collimator response correction (increases of 0.12 and 0.15 for acquired and simulated data, respectively, after FDP and attenuation corrections compared with attenuation correction alone) and scatter correction (increases of 0.09 and 0.10 for acquired and simulated data, respectively, after JAS and attenuation corrections compared with attenuation correction alone). Spatial resolution was also improved primarily by collimator response correction (3- and ~4-mm improvement in acquired and simulated data, respectively) and to a lesser extent by scatter correction (improvement < 1 mm in acquired and simulated data). Reconstruction using I20 + FBP gave comparable SNR values for simulated (4.9) and acquired (4.7) data. Attenuation correction slightly improved the SNR in both simulated and acquired data. On the other hand, collimator response and scatter corrections severely deteriorated the SNR: The SNR decreased by 1.2 and 1.8 after FDP correction for the acquired and simulated data, respectively (comparing I20 + MR with I20 + FDP + MR), whereas the SNR decreased by 0.8 for both acquired and simulated data after scatter correction (comparing I20 + MR with JAS + MR).

For all combinations of corrections, the quantitative accuracy of the LV wall and liver activity estimates was very close using the simulated and the acquired data (differences of ≤3%).

**Study Limitations**

Our experimental conditions differ in some respects from those encountered in clinical cardiac SPECT. We discuss here the simplifications we made—namely, the absence of patient motion and of background activity from surrounding organs and the use of an ideal attenuation map.

**Patient Motion.** In cardiac SPECT, motion is associated with heart contraction, the diaphragm motion caused by breathing, and, potentially, patient motion. As the wall thickness varies during the cardiac cycle, so does the underestimation associated with FSR. This was illustrated by Eisner et al. (38), whose dog studies showed that abnormal segmental contraction (with no alteration of perfusion) could result in apparent perfusion defects. Patient motion can also result in misregistrations between the emission and transmission scans. The effect of such misregistrations can be significant in cardiac SPECT. For instance, a 7-mm misregistration results in a modification of regional activity by 17% in the LV wall (39,40), and a 2- or 3-cm misregistration can produce nonuniformity of the LV activity distribution comparable with those observed without attenuation correction.

**Background Activity.** In our experiments, activity was simulated only in the LV wall and in the liver. In patients, nonspecific activity can also be observed in the LV cavity, in the lungs, or in the digestive tract. The activity concentration ratio between the liver and the LV wall that we considered was among those that can be observed in a healthy subject (0.60). However, this ratio can be closer or even higher than 1, particularly in patients with low cardiac uptake. Background activity affects image quality as well as quantitation accuracy: For instance, an activity concentration ratio of 2 between the liver and the LV can result in an increase of 10% of counts detected in the BEM (41). Therefore, our results should be considered as representative of the respective roles of the corrections, but further investigations are needed to determine the variability of our results for different imaging configurations.

**Attenuation Map.** For the real data, an acquired attenuation map was used; for the simulations, the ideal attenuation map was used. Although the 2 attenuation maps did not have the same spatial resolution, all evaluation criteria were in close agreement for simulated and acquired data (Table 2). The small impact of attenuation map resolution on BEM uniformity is in agreement with findings of Luo et al. (29) for numeric simulations of 201TI cardiac SPECT: These authors report a variation of <5% of the BEM uniformity index we used, after convolving the ideal attenuation map with a Gaussian of FWHM = 4.7 cm.

For our acquired data, transmission acquisition was performed using 2 153Gd line sources, and attenuation coefficients were linearly scaled from 100 to 140 keV. This scaling did not introduce a substantial bias given the good agreement observed between acquired and simulated data, with the latter using exact attenuation coefficients. This is in agreement with the 3% variation of BEM uniformity reported by Luo et al. (29) when attenuation coefficients were underestimated (or overestimated) by 30% in the case of 201TI cardiac SPECT.

**CONCLUSION**

Ideally, corrections should be made for scatter, attenuation, depth-dependent collimator response, or FSR in cardiac SPECT. We have shown that some corrections affect specific image characteristics (e.g., contrast, spatial resolution) more than others and have characterized quantitatively the respective impact of each correction for different features of interest. We also showed that some correction methods yielded results close to those that would be expected for ideal corrections but that the variability of the results obtained with different processing procedures increased with the number of corrections that were performed. In the simplified cardiac configuration considered in this study, we found that liver and LV wall activity could be estimated with errors < 10% using appropriate corrections. Although such accuracy will definitely be more difficult to achieve in vivo, our results suggest that accurate activity quantitation becomes possible in SPECT when adequate corrections are performed.

**ACKNOWLEDGMENTS**

The authors thank Dr. Stephen C. Moore for his helpful comments. This work was supported in part by grants from
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