Effects of motion, attenuation, and scatter corrections on gated cardiac SPECT reconstruction

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Purpose: In gated cardiac single photon emission computed tomography (SPECT), image reconstruction is often hampered by various degrading factors including depth-dependent spatial blurring, attenuation, scatter, motion blurring, and low data counts. Consequently, there has been significant development in image reconstruction methods for improving the quality of reconstructed images. The goal of this work is to investigate how these degrading factors will impact the reconstructed myocardium when different reconstruction methods are used.

Methods: The authors conduct a comparative study of the effects of these degrading factors on the accuracy of myocardium by several reconstruction algorithms, including (1) a clinical spatiotemporal processing method, (2) maximum likelihood (ML) estimation, (3) 3D maximum *a posteriori* (MAP) estimation, (4) 3D MAP with posttemporal filtering, and (5) motion-compensated spatiotemporal (4D) reconstruction. To quantify the reconstruction results, the authors use the following measures on different aspects of the myocardium: (1) overall error level in the myocardium, (2) regional accuracy of the left ventricle (LV) wall, (3) uniformity of the LV, (4) accuracy of regional time activity curves by normalized cross-correlation coefficient, and (5) perfusion defect detectability. The authors also assess the effectiveness of degrading corrections in reconstruction by considering an upper bound for each reconstruction method, which represents what would be achieved by each method if the acquired data were free from attenuation and scatter degradations. In the experiments the authors use Monte Carlo simulated cardiac gated SPECT imaging based on the 4D NURBS-based cardiac-torso (NCAT) phantom with different patient geometry and lesion settings, in which the simulated ground truth is known for the purpose of quantitative evaluation.

Results: The results demonstrate that use of temporal processing in reconstruction (Methods 1, 4, and 5 above) can greatly improve the reconstructed myocardium in terms of both error level and perfusion defect detection. In low-count gated studies, it can have even greater impact than other degrading factors. Both attenuation and scatter corrections can lead to reduced error levels in the myocardium in all methods; in particular, with 4D the bias can be reduced by as much as four-fold compared to no correction. There is a slight increase in noise level observed with scatter correction. A significant improvement in heart wall appearance is demonstrated in reconstruction results from three sets of clinical acquisitions as correction for degradations is combined with refinement of temporal filtering.

Conclusions: Correction for degrading factors such as resolution, attenuation, scatter, and motion blur can all lead to improved image quality in cardiac gated SPECT reconstruction. However, their effectiveness could also vary with the reconstruction algorithms used. Both attenuation and scatter corrections can effectively reduce the bias level of the reconstructed LV wall, though scatter correction is also observed to increase the variance level. Use of temporal processing in reconstruction can have greater impact on the accuracy of the myocardium than correction of other degrading factors. Overall, use of degrading corrections in 4D reconstruction is shown to be most effective for improving both reconstruction accuracy of the myocardium and detectability of perfusion defects in gated images. © 2011 American Association of Physicists in Medicine. [DOI: 10.1118/1.3660328]

Key words: gated cardiac SPECT, 4D reconstruction, motion compensation, attenuation compensation, scatter compensation

I. INTRODUCTION

Cardiac single photon emission computed tomography (SPECT) is a widely used, cost-effective imaging modality for assessment of cardiac artery diseases. The data acquisition is synchronized to the ECG signal in gated SPECT, which can provide valuable functional information such as ejection fraction, wall motion, and wall thickening. Cardiac SPECT images often suffer from a number of degrading factors, including attenuation, scatter, and depth-dependent spatial blurring; a gated study also suffers from an increased noise level due to further division of the acquisition up into a number of gates spanning the cardiac cycle.

Because of their importance, there have been significant efforts in development of correction methods for these degrading factors. Notably, for noise reduction in gated cardiac SPECT, there have been different image reconstruction methods developed, which include prefiltering, postfiltering, or joint spatiotemporal reconstruction methods, e.g., Refs. 1–5. A common theme in these methods is to enforce smoothing either spatially within a gated frame, or temporally across the different gated frames, or both. For brevity, these citations are rather limited as examples of different methods for noise reduction and are not intended to be a comprehensive literature review. In particular, in Ref. 2, a space-time Gibbs prior was defined as an extension of the conventional spatial Gibbs prior, in which an *a priori* motion model was used to adjust the weighting of time-domain cliques for temporal smoothing. It was demonstrated that such a prior could be more effective for noise smoothing than simple filtering. In Ref. 3, a simultaneous image reconstruction and motion estimation procedure was proposed for two image frames, in which a motion estimation term was included in the objective function along with a biomechanical regularization model on the motion field. It was found that the simultaneous method could reduce the error level compared with reconstruction without motion estimation. In Ref. 5, a 4D reconstruction approach was proposed, in which motion compensation was used to enforce temporal smoothing and the optical flow algorithm was used to determine the motion term. It was demonstrated that reconstruction with motion compensation could improve both the reconstruction accuracy and defect detectability.

Besides the noise effect, there also exists a body of work on how to correct for other factors such as attenuation, scatter, and spatial resolution in SPECT imaging (a rather comprehensive review of different correction techniques can be found in Ref. 6; below, the authors cite only a few example studies in the interest of brevity). If not properly accounted for, these factors are known to cause decreased contrast⁷ or even spurious perfusion defects⁸ in the reconstructed images. It is now generally accepted that attenuation and scatter corrections are necessary for quantitative imaging. However, there is no consensus yet on their benefit for improving lesion detection and sometimes even conflicting results have been reported in different studies.^{9–14} It is noted that these studies were mostly conducted using the summed images in gated cardiac imaging (owing to the high noise level in

individual gated frames) to assess perfusion lesions,^{9–12} and thus, the effect of cardiac motion was not considered. Furthermore, most of these studies were based on a particular reconstruction method.

The purpose of this investigation is to provide a comparative study of the effects of these degrading factors on the reconstruction of gated cardiac SPECT when different reconstruction algorithms are used. Such a study would allow us to examine the coupling effects between the degrading correction strategy and the reconstruction methods. It would also have considerable pragmatic value, because the obtained results would enable one to decide where best to invest efforts to improve reconstructing gated myocardial perfusion images. For example, will these degrading factors impact the different reconstruction methods in the same way? Is there any further room for improvement in a correction strategy? And, if given the choice, which degrading factor correction has the most impact on the resulting images for a particular reconstruction method? Moreover, for a given task, what is the best reconstruction algorithm and correction strategy? Our study will focus on the accuracy of the reconstructed myocardium in the gated frames. To our best knowledge, such a systematic, comparative study has not been reported yet in the literature.

We consider the following reconstruction methods of differing complexity: (1) a clinical spatiotemporal processing method, where gated frames are first reconstructed by filtered backprojection (FBP), followed by both spatial and temporal filtering, (2) maximum likelihood (ML) reconstruction, (3) 3D maximum *a posteriori* (MAP) reconstruction with a spatial smoothing prior, (4) 3D MAP reconstruction with post-temporal filtering, and (5) motion-compensated spatiotemporal (4D) reconstruction. Note that the 4D method previously developed in Ref. 5 is included here for consideration, as it was demonstrated to outperform several other methods. The results from this study are aimed to provide better understanding of the effects of different degrading factors on the various algorithms (4D included), which was not available in the previous work.⁵

To thoroughly evaluate the effects of the degrading factors on the reconstruction by the different methods, we consider the following quantitative measures: (1) overall error level in the reconstructed myocardium, (2) regional accuracy of the left ventricle (LV) wall, (3) uniformity of the LV wall, (4) accuracy of regional time activity curves (TACs) of LV by normalized cross-correlation coefficient, and (5) lesion detectability (i.e., diagnostic accuracy) in the reconstructed images quantified by a channelized Hotelling observer (CHO).^{15,16} With these measures, we also assess the effectiveness of degrading corrections in reconstruction by considering an upper bound for each reconstruction method, which represents what would be achieved by each method if the acquired data were free from degradations.

In our experiments, we use Monte Carlo simulation¹⁷ to simulate acquisition with the NURBS-based cardiac-torso (NCAT) phantom¹⁸ and Tc99m-Sestamibi as the imaging agent, based on which multiple noise realizations are obtained. We investigate two sets of NCAT phantom data

with different patient geometry, one male and one female; for lesion detection, different perfusion defects are introduced in the phantoms. The female phantom is set to have much higher breast and diaphragmatic attenuation than the male one. The idea is to demonstrate whether similar conclusions can be drawn with these two very different geometry settings. As a preliminary demonstration, we also present results by the different reconstruction methods from several sets of clinical acquisitions.

The rest of the paper is organized as follows: the reconstruction algorithms, evaluation criteria, and image data are described in Sec. II. Experiment results and discussions are given in Sec. III. Conclusions are given in Sec. IV.

II. METHODS

II.A. Reconstruction methods

Below, we briefly describe the different reconstruction methods used in this study. We will begin with a noniterative, clinical method, followed by iterative methods in the order of increased complexity.

- (1) *Clinical Spatiotemporal Method*: For comparison, we consider a clinical spatiotemporal processing method.⁵ In this method, the different gated frames are first reconstructed by FBP, then processed spatially with a Butterworth filter of order 2.4 and cutoff frequency of 0.2 cycles/voxel, and further processed along the gate dimension with a temporal filter with impulse response 1/4, 2/4, 1/4. This method is referred to as ST121.
- (2) *ML Reconstruction*: In this method, the different gated frames are reconstructed separately by ML estimation.¹⁹ Specifically, with f_k , $k=1, \dots, K$, denoting the gated frames, and g_k , $k=1, \dots, K$, denoting their corresponding acquired data, the ML estimate is obtained as follows:

$$\hat{f}_k = \max_{f_k} \log p(g_k | f_k), \quad k = 1, \dots, K, \quad (1)$$

where $p(g_k | f_k)$ is the likelihood function of g_k parameterized by f_k .

- (3) *3D MAP Reconstruction*: In this method a spatial smoothing prior is introduced to complement the image data. The MAP estimate is obtained as

$$\hat{f}_k = \max_{f_k} [\log p(g_k | f_k) + \log p(f_k)], \quad k = 1, \dots, K, \quad (2)$$

where $p(f_k)$ is a prior distribution on f_k . In this study a Gibbs prior of the following form is used: $p(f_k) \propto \exp[-\beta_s U_s(f_k)]$, where $U_s(f_k)$ is used to enforce spatial smoothing. For 3D volume, it is defined as the sum of the squared difference of each voxel from each of its 26 immediate neighbors. This method is referred to as MAP-S.

- (4) *3D MAP With Post-Temporal Filtering*: Here, the gated frames are first reconstructed individually with MAP-S, then further processed by the temporal filter as in ST121 above. This method is referred to as MAP-S + T121.

- (5) *4D Reconstruction*: We consider a motion-compensated four-dimensional (4D) reconstruction method recently developed in our previous work,⁵ in which the different gated frames are reconstructed simultaneously through MAP estimation. Briefly, let $\mathbf{F} = (f_1^T, f_2^T, \dots, f_K^T)^T$ denote the collection of the gated frames and $\mathbf{G} = (g_1^T, g_2^T, \dots, g_K^T)^T$ denote their corresponding acquired data. The gated frames are reconstructed jointly by using MAP estimation as follows:

$$\hat{\mathbf{F}} = \max_{\mathbf{F}} [\log p(\mathbf{G} | \mathbf{F}) + \log p(\mathbf{F})], \quad (3)$$

where $\log p(\mathbf{G} | \mathbf{F})$ is the likelihood function of $\mathbf{G}$ parameterized by $\mathbf{F}$, and $p(\mathbf{F})$ is a prior distribution on $\mathbf{F}$.

In Eq. (3), the prior term $p(\mathbf{F})$ is used to enforce spatiotemporal regularization. It is defined in the form $p(\mathbf{F}) \propto \exp[-\beta_s U_s(\mathbf{F}) - \beta_t U_t(\mathbf{F})]$, where $U_s(\mathbf{F})$ is used to enforce spatial smoothing within a gated frame, and $U_t(\mathbf{F})$ is used to exploit temporal smoothing among the gated frames. Specifically, the term $U_s(\mathbf{F})$ consists of the differences of each voxel from its 26 spatial neighbors in a gated frame; likewise, the term $U_t(\mathbf{F})$ consists of the motion-prediction errors of a gated frame from its temporal neighboring frames. The effects of these two terms are controlled through their respective weighting parameters β_s and β_t .

In our experiments, the technique of ordered subsets was used to speed up the iterative methods above. Specifically, the OSEM algorithm²⁰ was used for ML reconstruction (1), and the BSREM algorithm was used in both 3D MAP and 4D.⁵ For these algorithms, a total of 16 subsets and 10 iterations were used. This choice was determined empirically in our previous studies.⁵

II.B. Compensation of degrading factors

To compensate for degrading effects in reconstruction, we have included several factors directly in the forward imaging model, which relates the unknown images to the sinogram data. Specifically, (1) the depth-dependent blur is modeled in the system matrix; (2) the attenuation effect is also included in the system matrix; and (3) for scatter correction, the scatter component is embedded in the likelihood function; the filtered TEW method⁶ was used to estimate the scatter components in our experiments. More details of these compensations can be found in Ref. 5.

In our experiments, we thoroughly tested the effects of various degrading factors on the reconstruction. For comparison, we present results obtained with different reconstruction methods for each of the following cases: (1) no attenuation and scatter correction (NC); (2) attenuation correction only (AC); (3) both attenuation and scatter correction (ASC), and (4) reconstruction from acquisition without attenuation and scatter (Ideal-ASC). Note that, in the above, Ideal-ASC is used as an upper bound, which represents what would be achieved by each method if the acquired data were free from attenuation and scatter. This bound is used to evaluate the effectiveness of degrading corrections in the different reconstruction methods.

To summarize, in Table I, we show how the different correction methods were evaluated for the different reconstruction methods. For example, temporal processing was included in the following reconstruction methods: ST121, MAP+T121, and 4D. Note that the depth-dependent blur was corrected in all the reconstruction methods except for ST121. Thus, ST121 serves as a baseline for no resolution correction.

II.C. Quantification of reconstruction

To evaluate the effects of the degrading factors on the reconstructed images, we compute a number of quantitative metrics, including: (1) overall error level of the myocardium, (2) regional bias-variance analysis of the LV wall, (3) uniformity of the LV wall, (4) TAC analysis of LV wall, and (5) detectability of perfusion defects. These metrics are used to quantify several different aspects of the reconstructed myocardium, as defined below.

- (1) *Overall Error Level of Myocardium*: The overall reconstruction error level in the myocardium is measured by mean square error (MSE) as

$$\text{MSE} = \frac{1}{NK} \sum_{k=1}^K \|f_k - \hat{f}_k\|^2, \quad (4)$$

where f_k and $\hat{f}_k$ denote the reference and reconstructed images of gate k over a $30 \times 28 \times 20$ volumetric region containing the entire LV, of which a 2D slice is shown in Fig. 2, and N is the total number of voxels within the volumetric region. Note that the MSE defined above is computed for the myocardium ROI over all the K gated frames.

- (2) *LV Regional Bias-Variance Analyses*: To quantify the regional accuracy of LV, we conduct bias-variance analyses on regions of interest (ROIs) selected on the LV wall as shown in Figs. 2(a) and 2(b). Specifically, let $\bar{f}_{\text{ROI}}$ denote the mean intensity of the ROI in the noiseless reference f , i.e.,

$$\bar{f}_{\text{ROI}} = \frac{1}{M} \sum_{i \in \text{ROI}} f(i), \quad (5)$$

TABLE I. Correction of degrading factors in different reconstruction methods: Depth-dependent blur correction (Blur), no attenuation and scatter correction (NC), attenuation correction (AC), both attenuation and scatter correction (ASC), and temporal processing (temporal). Note that depth-dependent blur is corrected in all the reconstruction methods except for ST121, and temporal processing is included in all the methods except for ML and MAP-S.

Methods	Blur	NC	AC	ASC	Temporal
ML	Yes	Yes	Yes	Yes	No
MAP-S	Yes	Yes	Yes	Yes	No
MAP-S + T121	Yes	Yes	Yes	Yes	Yes
4D	Yes	Yes	Yes	Yes	Yes
ST121	No	No	No	No	Yes

where M is the total number of voxels within the ROI. Our goal is to quantify the statistical accuracy of this quantity from different noisy reconstructions.

Let $\hat{f}^{(q)}$, $q = 1, 2, \dots, Q$, denote estimates of f obtained from Q different noise realizations. The mean estimate is computed as

$$\hat{\mu} = \frac{1}{Q} \sum_{q=1}^Q \bar{f}_{\text{ROI}}^{(q)}. \quad (6)$$

Then, the bias and standard deviation (std) of this mean estimator are, respectively, estimated as

$$\text{bias} = \frac{|\bar{f}_{\text{ROI}} - \hat{\mu}|}{\bar{f}_{\text{ROI}}} \times 100\%, \quad (7)$$

and

$$\text{std} = \frac{\sqrt{\frac{1}{Q-1} \sum_{q=1}^Q (\bar{f}_{\text{ROI}}^{(q)} - \hat{\mu})^2}}{\bar{f}_{\text{ROI}}} \times 100\%. \quad (8)$$

In our experiments, a total of $Q = 30$ noise realizations were used.

- (3) *LV Uniformity*: We quantify the image uniformity around the LV wall of a normal study in short-axis view. Specifically, a set of 32 voxels are selected along the heart wall in one short-axis slice [Figs. 2(c) and 2(d)]; the mean of the intensity values at this set of voxels is first computed, and their average deviation from this mean is then computed as a measure of the wall uniformity. That is, the wall uniformity is defined as

$$U = \frac{\sum_{i=1}^{M'} |f_{\text{ROI}}(i) - \bar{f}_{\text{ROI}}|}{M' \bar{f}_{\text{ROI}}} \times 100\%, \quad (9)$$

where $\{f_{\text{ROI}}(i), i = 1, \dots, M'\}$ denotes the selected set of voxels on the heart wall, and $\bar{f}_{\text{ROI}}$ is the mean value of this set of voxels.

- (4) *TAC*: In order to demonstrate the effect of potential temporal smoothing on cardiac motion, we also calculate the TAC of an ROI near the base of the LV myocardium shown in Figs. 2(c) and 2(d). As the wall moves in and out of this ROI during the beating cycle, the intensity of the ROI will vary accordingly and thus serves as a good indicator on the degree of temporal smoothing.

The normalized cross-correlation coefficient (CC) between a reconstructed TAC and its ideal reference is computed as follows:

$$\text{CC}(\mathbf{y}, \hat{\mathbf{y}}) = \frac{(\mathbf{y} - \bar{\mathbf{y}})^T (\hat{\mathbf{y}} - \bar{\hat{\mathbf{y}}})}{\|\mathbf{y} - \bar{\mathbf{y}}\| \cdot \|\hat{\mathbf{y}} - \bar{\hat{\mathbf{y}}}\|}, \quad (10)$$

where $\mathbf{y} = [y(1), \dots, y(K)]^T$ and $\hat{\mathbf{y}} = [\hat{y}(1), \dots, \hat{y}(K)]^T$ are the ideal and reconstructed TACs of selected ROI; $\bar{\mathbf{y}}$ and $\bar{\hat{\mathbf{y}}}$ are the mean values of $\mathbf{y}$ and $\hat{\mathbf{y}}$, respectively.

(5) *Lesion Detectability*: Furthermore, to quantify lesion detectability in the reconstructed images, which is relevant to diagnostic accuracy, a CHO is used. In the CHO, four rotationally symmetric, nonoverlapping input channels are used and internal noise is included as in Ref. 5. A total of 60 noise realizations (30 lesion-present and 30 lesion-absent) are used and the reconstructed images by each method are assessed by the CHO. The detection performance is summarized using the area under the receiver operating characteristic (ROC) curve. These ROC studies represented a “signal-known exactly” and “background-known exactly” observer study.

As noted in the introduction, the effects of degrading factors on lesion detection are mostly conducted using ungated images in the literature because of the high noise level in conventional gated images. However, this also ignores the cardiac motion which could potentially cause blurring of the defect. In this study, we will investigate whether the different methods could potentially lead to improved detectability of perfusion defects in individual gated frames. With improved noise suppression, it may open the door to study the feasibility of using gated images for defect detection, which is certainly an interesting topic worthy of further clinical investigation. Toward this purpose, the CHO is used to evaluate the lesion detectability in individual gated frames of the reconstruction. In our experiments, the CHO results were obtained for the first gated (end-diastole) frame during which the heart wall is expanded and small lesions are easier to be seen.

II.D. Image data

(1) *Phantom Simulation Data*: In our experiments, the 4D NCAT 2.0 phantom¹⁸ was used to generate the source and attenuation distribution. The SIMIND Monte Carlo package¹⁷ was used to simulate gated SPECT imaging with Tc-99m labeled Sestamibi as the imaging agent and with the Picker Prism3000 with low-energy high-resolution (LEHR) collimator as the imaging system. The projection matrix was 64 by 64 with a pixel size 0.634 cm. For a circular camera rotation of 28.5 cm radius, 64 pro-

jection sets were collected for each gated frame of total 16 gates. Two energy windows were used as in Ref. 5: the photopeak window was 20% (28 keV) centered at 140 keV, and a 3.5 keV window abutted to the lower side of the photopeak window for scatter correction. The scattered counts in the photopeak window were about 34% of the total counts. Poisson noise was introduced at a level of 8×10^6 total acquired counts as in a typical clinical acquisition for a Tc-99m labeled Sestamibi study.

Two versions of the phantom, one male and one female, were generated; one transverse slice is shown in Fig. 1 for each phantom. The male phantom (referred to as Phantom A hereinafter) was generated with 35.2 cm body long-axis size, 26.7 cm body short-axis size, 41.7 cm body height, 30 cm rib long-axis size, 22.7 cm rib short-axis size and 37.3 cm rib height. The female phantom (referred to as Phantom B hereinafter) was generated with the same 35.2 cm body long-axis size, 26.7 cm body short-axis size, 41.7 cm body height, but a smaller 24 cm rib long-axis size, 18.2 cm rib short-axis size and 29.8 cm rib height. The simulated breast dimensions were 18.2, 7.0, and 14.0 cm in long-axis, short-axis and height, respectively. Consequently, Phantom B suffers from more significant breast and diaphragmatic attenuation.

For each phantom, two datasets were generated, one with a lesion and the other without. For Phantom A, a transmural perfusion defect with 20% intensity reduction was introduced in the septal wall of the LV. For Phantom B, a transmural perfusion defect with 50% intensity reduction was introduced in the anterior-lateral region of the LV. These lesion defect locations are shown in Figs. 2(a) and 2(b). Each dataset included 30 noise realizations. There is also a high concentration of activities in the liver. For attenuation correction, the corresponding attenuation map from each NCAT phantom was used.

As reference for quantitative analyses, the gated volumetric images were reconstructed using the OSEM algorithm (10 iterations with 16 subsets) from simulated noiseless projection of the phantom data without attenuation and scatter (by turning off attenuation and scatter in SIMIND); depth-dependent blurring was included both

FIG. 1. Transverse slices of male and female phantoms (Phantoms A and B, respectively).

FIG. 2. Regions of interest (ROIs) defined in the two phantoms for bias-variance, CHO, TAC and uniformity analyses. These are from original source distributions without any degradations; (a) and (b) are shown in transverse slices, and (c) and (d) are in short-axis slices. Note that (a) and (b) are shown with interpolation for CHO.

in the simulated data and in the reconstruction algorithm. This corresponds to the ideal case of perfect acquisition where none of attenuation, scattering, and Poisson noise are involved in the sinogram data (denoted as “Ideal” below for reference). This ideal case represents what would be achievable if attenuation, scatter and noise were not present. By comparing to this noiseless reconstruction, we will learn how far the reconstructed images are from this ideal situation.

Since imaging noise is inevitable in practice, we also considered a somewhat more “realistic” reference for each reconstruction method, where we simulated data acquisitions with no attenuation and scatter as in the Ideal case above except that noisy projections were used. As described earlier in Sec. II B (referred to as Ideal-ASC), this represents what would be achieved if both attenuation and scatter degrading factors were not present in the acquired data. For comparable performance, the acquired count level on average was kept the same as that of the primary photons in simulations above with both attenuation and scatter.

- (2) *Clinical Data:* As a preliminary demonstration, we also used several sets of clinical data. These data sets were acquired from patients by an IRIX system²¹ with 68 projections (three-degree steps) and a 128×128 matrix. The pixel size was 0.467 cm. The acquisition started from right anterior oblique, passed through anterior and left anterior oblique, and ended at left lateral oblique. A total of eight gates were used. The photopeak window was 15% centered at 147.5 keV, and a second 5% window centered at 133.2 keV was used. The total number of counts acquired was 5.57×10^6 , 8.21×10^6 , and

11.03×10^6 , respectively. For attenuation correction, the attenuation maps were estimated from the transmission images of the patients. Note that in these sets of clinical data the photopeak window was set to be centered at 147.5 keV so that it was asymmetric to the high side in order to decrease scatter within the acquired photopeak window data.²²

III. RESULTS AND DISCUSSIONS

III.A. Overall error level of myocardium

We show in Fig. 3 the MSE results of the reconstructed myocardium with different reconstruction methods under different degrading corrections. Specifically, in Fig. 3(a), we show together the results obtained from both ML and 3D MAP-S reconstructions; each curve in the plot was obtained by varying the spatial parameter β_s for 3D MAP-S (abscissa). (ML corresponds to $\beta_s = 0$). Similarly, in Fig. 3(b), we show the results obtained from 3D MAP-S with post-temporal filtering. In Fig. 3(c), we show the results from 4D reconstruction; the optimal parameter value β_t was used in each setting (Ideal-ASC: $\beta_t = 0.01$; ASC: $\beta_t = 3 \times 10^{-4}$; AC: $\beta_t = 2 \times 10^{-4}$; NC: $\beta_t = 2 \times 10^{-3}$). All these results were averaged over 30 noise realizations. For fair comparison, the reconstructed images from the different methods were all rescaled to have the same count level as that of the acquired data in each realization; this is of consideration that the count levels in the reconstructed images would vary with the degrading correction methods applied. This scaling step will greatly reduce the apparent (large) bias associated with both attenuation and scatter when they were not corrected for in the reconstruction.

FIG. 3. Mean square error (MSE) of reconstructed myocardium with different reconstruction and degrading correction methods for Phantom A: (a) ML and 3D MAP reconstruction (MAP-S), (b) 3D MAP with post-temporal filtering (MAP-S + T121), (c) 4D. Note that a smaller MSE value indicates a better agreement with the reference.

From these results, it can be seen that for reconstruction with methods MAP-S, MAP-S + T121 and 4D, the best MSE was all obtained with ASC, followed by AC and NC; compared to NC, attenuation correction resulted in most of the improvement, and use of scatter correction could lead to further improvement (albeit small in MAP-S). For ML reconstruction, however, the best MSE was obtained with AC; additional use of scatter correction actually increased MSE. As will be seen from subsequent results, this is due to increased variance level in the reconstruction associated with scatter correction.

Interestingly, from Fig. 3(a), it can be seen that the MSE of ST121 reconstruction is even comparable to the best MSE achieved by MAP-S with AC or ASC. On the surface, this might seem rather surprising given that no corrections were done for resolution, attenuation, and scatter in ST121.

However, this demonstrates that temporal smoothing could be more effective for noise reduction than other degrading factors. Indeed, with similar temporal processing included in MAP-S + T121, the results in Fig. 3(b) show the additional benefit of AC and ASC. In addition, among the different reconstruction methods, the best MSE was obtained by 4D with $\beta_s = 0$ (i.e., without spatial smoothing); with ASC, the MSE from 4D was close to the upper bound from “Ideal-ASC.” These results show that temporal smoothing (4D, MAP-S + T121, ST121) could be more accurate than spatial smoothing in reconstruction.

Similar MSE results were also obtained for Phantom B, where the relative performance by the different reconstruction methods was consistent with that of Phantom A. However, the MSE level in Phantom B was observed to be much higher than that in Phantom A, thanks to the significantly

higher attenuation in B. Interestingly, the results show that the effects of AC and ASC were also greater in Phantom B than in Phantom A. For brevity, these results are not shown here.

III.B. LV regional bias-variance

We conducted bias-variance analyses on two ROIs (one normal and one defect) on the LV wall in both the male and female phantoms (Fig. 2). We show in Fig. 4 the bias-standard deviation plots of the normal ROI obtained with the different reconstruction methods under different degrading corrections (ASC, AC, or NC). In these plots, each curve was obtained by varying the spatial parameter for which the same range was used as in Fig. 3: $\beta_s = [0, 0.5, 2, 5, 7] \times 10^{-4}$ for ASC, $\beta_s = [0, 0.5, 2, 3, 5] \times 10^{-4}$ for AC, $\beta_s = [0, 0.5, 1, 5, 8] \times 10^{-2}$ for NC and $\beta_s = [0, 0.5, 1, 3, 5] \times 10^{-2}$ for Ideal-ASC. In each curve, the left-end point corresponds to $\beta_s = 0$ (i.e., no spatial smoothing); thus, each curve in Fig. 4 shows a decreasing trend in variance but also an increasing trend in bias as the spatial parameter β_s is increased. As above, all these results were obtained from 30 noise realizations.

From these results, it can be seen that in all reconstruction methods the use of degrading corrections could lead to significant reductions in the bias level of the LV. For example, in 4D reconstruction of Phantom A, the bias level with AC was reduced to $\sim 25\%$ from the level of $\sim 35\%$ for NC; with additional scatter correction (ASC), the bias could be further reduced down to below 10%, approximately a four-fold reduction compared to NC. Interestingly, with a substantial reduction in bias, ASC also led to a slight increase in variance compared to AC in all reconstruction methods. Similar observations can be made from the results of Phantom B; in particular, the bias level in Phantom B is noted to be higher than that in Phantom A. This is caused by the more pronounced attenuation effects in Phantom B.

It is also observed from Fig. 4 that use of temporal processing could greatly decrease the variance in the reconstruction. Compared to MAP-S, MAP-S + T121 could achieve much smaller variance for all compensation methods, and 4D could yield even further reduction in variance. Moreover, while spatial smoothing can be effective for noise reduction (i.e., decreasing the variance level), it also can lead to spatial blurring of the heart wall (i.e., increasing the bias level). However, with motion-compensated temporal smoothing in 4D, the benefit of spatial smoothing diminishes. Indeed, the results earlier in Fig. 3 show that the best MSE was obtained by 4D with $\beta_s = 0$. As demonstrated subsequently, this improved accuracy of the heart wall can also lead to better detection of perfusion defects.

Similar results were also obtained for the defect ROI in both phantoms. For brevity, these results are not shown here.

III.C. LV uniformity

We show in Fig. 5 the uniformity results of the LV wall obtained by the different reconstruction methods under different degrading corrections (ASC, AC, or NC) with

Phantom A. These results were computed from the reconstructed images of 30 noise realizations by each method (the standard deviation values are indicated by the errorbars). For each reconstruction method, the value of β_s was selected based on the best MSE results in Fig. 3 for each degrading correction. Similar results were also obtained with Phantom B, but omitted here in the interest of brevity.

As can be seen, compared to NC, AC could lead to improved uniformity of the LV wall in reconstruction methods 4D and MAP-S + T121; in particular, 4D yielded the best uniformity of the LV wall (compared to Ideal). However, when additional scatter correction was used, the results showed reduced wall uniformity. While this might seem surprising at first sight, it is actually consistent with the regional bias-variance results of the LV earlier in Fig. 4, which showed an increase in variance in ASC compared to AC. This increased noise variance has impacted on the wall uniformity. For a similar reason, little or no improvement in wall uniformity was achieved by ML and MAP-S with AC or ASC.

III.D. TAC analyses

In Fig. 6, we show the normalized CC between the reconstructed and the ideal TACs of an ROI on LV [Fig. 2(c)] obtained by the different reconstruction methods under different degrading corrections (ASC, AC, NC, or Ideal-ASC) with Phantom A. As above, the parameter β_s was selected from the best MSE for each method. Similar results were also obtained with Phantom B.

These results show that 4D with AC could lead to the highest CC values in the reconstructed TACs (highest mean CC value and smallest variance); the improvement was also observed to be greater in Phantom B, which suffers from more attenuation. However, when additional scatter correction was used, the CC results even showed a slight reduction. This is consistent with the increased noise variance discussed above. In addition, it is also observed that, other than 4D, use of AC or ASC led to little or no improvement in CC results for all the other reconstruction methods.

III.E. Lesion detectability

In Fig. 7, we show the lesion detectability results (measured by the CHO) obtained by the different reconstruction methods under different degrading corrections (ASC, AC, NC, or Ideal-ASC), where the area under the ROC curve (Az) is shown. In these plots, each curve was obtained by varying the spatial parameter β_s (abscissa). For comparison, results are shown for both Phantoms A and B.

From Figs. 7(a) and 7(b), it can be seen that Az level achieved by both ML and MAP-S was similar to that of ST121. However, with additional temporal processing in MAP-S + T121, the Az level was notably improved in Figs. 7(c) and 7(d), particularly for Phantom B. In comparison, the best Az results were obtained by 4D in Figs. 7(e) and 7(f).

These results show that use of AC or ASC could improve the Az value in the reconstruction methods, with the best detectability obtained with ASC, followed by AC and NC.

(a) ML, MAP-S - Phantom A

(b) ML, MAP-S - Phantom B

(c) MAP-S+T121 - Phantom A

(d) MAP-S+T121 - Phantom B

(e) 4D - Phantom A

(f) 4D - Phantom B

FIG. 4. Bias-standard deviation plots of a normal ROI obtained with different reconstruction and degrading correction methods: (a) Phantom A with ML and 3D MAP reconstruction (MAP-S), (b) Phantom B with ML and MAP-S, (c) Phantom A with 3D MAP with post-temporal filtering (MAP-S + T121), (d) Phantom B with MAP-S + T121, (e) Phantom A with 4D, and (f) Phantom B with 4D. Note that the lower left corner in these curves corresponds to the lowest noise level and bias.

FIG. 5. Uniformity measure of the LV wall obtained with different reconstruction and degrading correction methods with Phantom A.

Such improvement is also most evident in the 4D results. For Phantom A, the best Az by 4D was as high as 0.856 with ASC, 0.816 with AC, and 0.771 with NC. For Phantom B, the effect of attenuation correction is more pronounced; the best Az value by 4D with AC alone was 0.929, almost as high as that of ASC (0.955). It is also noted that in nearly all cases the best Az values for 4D were obtained when there was little or no spatial smoothing applied ($\beta_s = 0$). This is also consistent with the MSE presented earlier.

Compared to 4D, the results in Fig. 7 also show that the improvement in Az was somewhat limited for reconstruction methods ML, MAP-S, or MAP-S + T121. Specifically, for

Phantom A, the best Az by MAP-S + T121 was 0.803 with ASC, 0.771 with AC, and 0.760 with NC; for Phantom B, the best Az by MAP-S + T121 was 0.826 with ASC, 0.817 with AC, and 0.799 with NC.

At first sight, the above results might seem somewhat surprising, because the quantitative results earlier (Figs. 3 and 4) clearly showed that both AC and ASC could also effectively improve the error level in reconstruction methods ML, MAP-S, or MAP-S + T121. Unfortunately, this error reduction may not effectively improve the detection of the perfusion defects. These results reveal that correcting for AC and ASC could improve lesion detectability; it can be most

FIG. 6. Normalized cross-correlation coefficients of reconstructed TACs obtained with different reconstruction and degrading correction methods with Phantom A.

FIG. 7. Channelized Hotelling observer performance obtained with different reconstruction and degrading correction methods: (a) Phantom A with ML and 3D MAP reconstruction (MAP-S), (b) Phantom B with ML and MAP-S, (c) Phantom A with 3D MAP with post-temporal filtering (MAP-S + T121), (d) Phantom B with MAP-S + T121, (e) Phantom A with 4D, and (f) Phantom B with 4D. Note that a larger A_z value indicates better detection accuracy.

FIG. 8. Reconstructed images of myocardium (Phantom B) in short-axis at end-diastole (ED) and end-systole (ES) by different methods.

beneficial when image motion is also taken into account in 4D reconstruction.

Finally, from Fig. 7, it is observed that the Az improvement is generally higher in Phantom B than in Phantom A by the different reconstruction and correction methods (e.g., AC vs. NC, or 4D vs. MAP-S + T121). We believe there are two main contributing factors to this. First, there is more significant breast and diaphragmatic attenuation in Phantom B than in Phantom A; thus, AC and ASC would lead to more error reduction in Phantom B, as demonstrated in the quantitative results with MSE and bias-variance analyses. Second, the higher contrast of the introduced defect in Phantom B

(50%, compared to 20% in Phantom A) also leads to higher detectability, as indicated by the upper bound Ideal-ASC in Fig. 7.

III.F. Reconstructed images

As an example, we show in Fig. 8 a set of reconstructed images of the myocardium of Phantom B from a typical noise realization by different methods. For brevity, images are shown for only two different gates [end-diastole (ED) and end-systole (ES)], and for each of the following reconstruction methods: ML, MAP-S, MAP-S + T121, 4D, and

FIG. 9. Reconstructed images of myocardium in short-axis at end-diastole (ED) and end-systole (ES) from three Tc-99m Mibi SPECT patient studies.

ST121; for each reconstruction method, results are shown for NC, AC, ASC, and Ideal-ASC. The noiseless reference images (Ideal) are also shown.

As can be seen, for all the reconstruction methods, the LV wall is notably more uniform in images obtained with AC than with NC; the images from temporal smoothing methods (4D and MAP-S + T121) suffer from significantly less noise artifacts compared to that of MAP-S. Furthermore, the images of 4D with ASC also show better definition of the LV wall.

III.G. Clinical images

Finally, we show in Fig. 9 short-axis slices at two different gates (ED and ES) by ML, MAP-S, MAP-S + T121, 4D, and ST121 from three patient studies. Owing to lack of ground truth, the parameters for MAP-S and 4D were chosen empirically, where more smoothing was applied for patient 1 due to its low counts (patient 1: $\beta_s = 5 \times 10^{-4}$ and $\beta_t = 5 \times 10^{-4}$; patients 2 and 3: $\beta_s = 1 \times 10^{-4}$ and $\beta_t = 1 \times 10^{-4}$). Note that in all three patients the wall shape is increasingly cleaner, and the mottled or noise blobby appearance of the walls is decreasing as one progresses from ML to 4D processing. For patients 1 and 3, notice the correction of apparent inferior wall decrease in uptake by ASC which is consistent with correction of attenuation of induced artifactual perfusion defects in this region; improvement is also observed in the lateral wall for patients 2 and 3.

From a cine display of these clinical images and phantom images, a notable improvement is observed in the smoothness of wall motion appearance as correction for degradations is combined with refinement of temporal filtering.

IV. CONCLUSIONS

In this work, we investigated the effects of resolution, attenuation, scatter, and motion correction in cardiac SPECT reconstruction with several different reconstruction algorithms, including a clinical spatiotemporal method (ST121), ML, 3D MAP (MAP-S), 3D MAP followed by temporal smoothing (MAP-S + T121), and joint spatiotemporal (4D) reconstruction. We quantified the reconstruction results using several metrics on different aspects of the reconstructed myocardium, ranging from overall error level, regional bias-variance analysis of the LV, uniformity of the LV wall, time activity curves, to lesion detectability. To assess the effectiveness of degrading correction in reconstruction, we also compared the results against a “realistic” upper bound which represents what would be achievable by each reconstruction method if the degrading factors were not present in the acquired data.

Our results demonstrate that use of both attenuation and scatter corrections can achieve the best MSE level in myocardium for reconstruction with MAP-S, MAP-S + T121, and 4D; a further bias-variance analysis shows that both attenuation and scatter corrections can effectively reduce the bias level of the reconstructed LV wall. However, while reducing the bias, scatter correction is also observed to increase the variance (i.e., noise) level. This increased noise

is noted to negate the reduction in bias in ML reconstruction, leading to increased MSE in the myocardium.

Furthermore, it is also observed that use of temporal processing in reconstruction (ST121, MAP-S + T121, and 4D) can greatly improve the reconstruction accuracy of the myocardium; it can even have greater impact than correction of other degrading factors. Remarkably, the clinical method ST121 is noted to achieve an MSE level that is even comparable to that of MAP-S with all resolution, attenuation, and scatter corrections. Interestingly, the lesion detection results show that ST121 could also achieve a similar detection level to that of ML and MAP-S.

In all reconstruction methods, attenuation correction is observed to improve the uniformity of the LV wall. It also leads to more accurate results in time activity curves. In contrast, scatter correction is observed to have negative effects on both wall uniformity and time activity curves, which are largely due to the increased noise in the reconstructed myocardium.

Overall, use of degrading corrections in 4D reconstruction is shown to be most effective for improving both reconstruction accuracy of the myocardium and detectability of perfusion defects. In particular, both attenuation and scatter corrections with 4D are observed to lead to the most improvement in lesion detectability than with any other methods. The improvement is also observed to be even greater in the presence of increased breast and diaphragmatic attenuation.

As a preliminary demonstration, we also tested the different reconstruction and correction methods on several sets of real patient data. These images are observed to be consistent with our quantitative results. In the future, we plan to conduct further evaluations using a significantly larger set of clinical acquisitions and expert observers.

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