Optimization of Ga-67 imaging for detection and estimation tasks: Dependence of imaging performance on spectral acquisition parameters

Georges El Fakhri,[a)] Stephen C. Moore, and Marie Foley Kijewski
Department of Radiology, Harvard Medical School and Brigham and Women's Hospital, Boston, Massachusetts 02115

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We have compared the use of two (93 and 185 keV) and three (93, 185, and 300 keV) photopeaks for Ga-67 tumor imaging and optimized the placement of each energy window. Methods: The bases for optimization and evaluation were ideal and Bayesian signal-to-noise ratios (SNR) for the detection of spheres embedded in a realistic anthropomorphic digital torso phantom and ideal SNR for the estimation of their size and activity concentration. Seven spheres of radii ranging from 1 to 3 cm, located at several sites in the torso, were simulated using a realistic Monte Carlo program. We also calculated the ideal SNR for the detection from simple phantom acquisitions. Results: For detection and estimation tasks, the optimum windows were identical for all sphere sizes and locations. For the 93 keV photopeak, the optimal window was 84–102 keV for the detection and 87–102 keV for estimation; these windows are narrower than the 20% window often used in the clinic (83–101 keV). For the 185 keV photopeak, the optimal window was 170–220 keV for the detection and 170–215 keV for estimation; these are substantially different than the 15% window used in our clinic (171–199 keV). For the 300 keV photopeak, the optimal window for detection was 270–320 keV, and for estimation, 280–320 keV. Using the three optimized, rather than only the two lower-energy, windows yielded a 9% increase in the SNR for the detection of the 3 cm diam sphere (a 12% increase for a 2 cm diam sphere) and a 7% increase in the SNR for estimation of its size. For the acquired phantom data, detection also increased by 9%–12% when using three, rather than two, energy windows.
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Key words: Ga-67 imaging, task-based optimization, detection, estimation, Bayesian estimator, Cramer–Rao bound

INTRODUCTION

Ga-67 SPECT is useful for the diagnosis and staging of non-Hodgkin's lymphoma, as well as of other cancers. In addition to the routine use of Ga-67 images for tumor detection and localization, estimates of tumor activity concentration from these images could also be useful for grading tumors, since gallium avidity in non-Hodgkin's lymphoma is correlated with the histopathologic grade of the tumor.[1,2] The clinical acquisition parameters, in particular, the number and placement of energy windows have not previously been optimized for these clinically relevant detection and estimation tasks. For example, data from only two of the three major Ga-67 photopeaks (93 and 185 keV) are often used (e.g., Bartold[2] and Wells[3]), although the inclusion of the 300 keV data could, potentially, improve lesion detectability. In this study we compared the use of data from two (93 and 185 keV) and three (93, 185, and 300 keV) photopeaks, and we optimized the upper and lower limits of each energy window. Because both detection and estimation are of interest, we based the optimization on metrics for both kinds of tasks. The detection task SNR was calculated using both a Bayesian observer, for which knowledge of the range of possible lesion sizes was incorporated as a prior,[4] and a nonprewhitening (NPW) signal-known-exactly/background-known-exactly (SKE/BKE) model. For the activity and size estimation tasks, we used figures of merit that we have previously proposed.[4] These are signal-to-noise ratios (SNR) consisting of the true value of the parameter of interest (in this case tumor activity or size) divided by the square root of the Cramer–Rao lower bound (CRB) on the variance with which the parameter can be estimated by an unbiased procedure.[5] These metrics were calculated using data obtained from realistic Monte Carlo (MC) simulations that modeled 72-hour post-injection Ga-67 studies. The simulations were based on a digital version of an anthropomorphic torso phantom, and all interactions in the patient, as well as in the detector, were modeled in the MC simulation. The detection metric was also calculated for physical acquisitions performed with the actual torso phantom under similar experimental conditions.

MATERIALS AND METHODS

Generation of digital anthropomorphic phantom

An anthropomorphic digital torso phantom that mimics an average male patient was created by segmenting the RSD torso phantom (Radiology Support Devices Inc., Long Beach, CA). The phantom was scanned on an x-ray CT scanner (Siemens SOMATOM) to obtain 156 transaxial slices, each 256×256, after injecting various concentrations of iodinated contrast agents into the different compartments.[6] The

Representative Results:

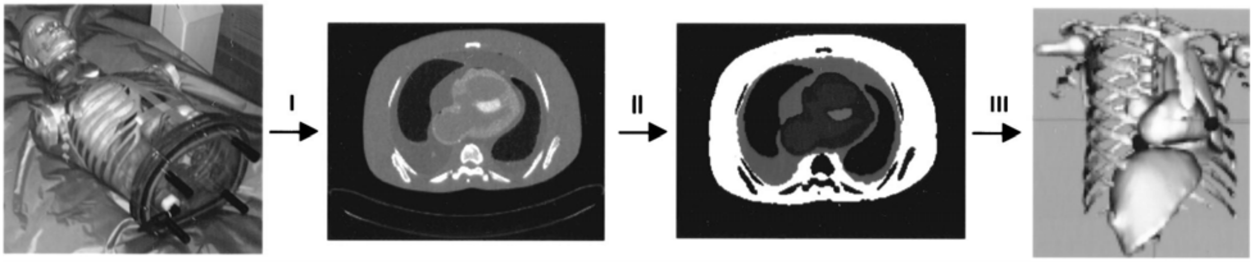

FIGURE 1. Generation of the digital anthropomorphic phantom: I: CT scan of the physical phantom; II: segmentation; III: surface rendering of the digitized torso phantom, with the right lung not shown for better visualization of the spine and the posterior tumor.

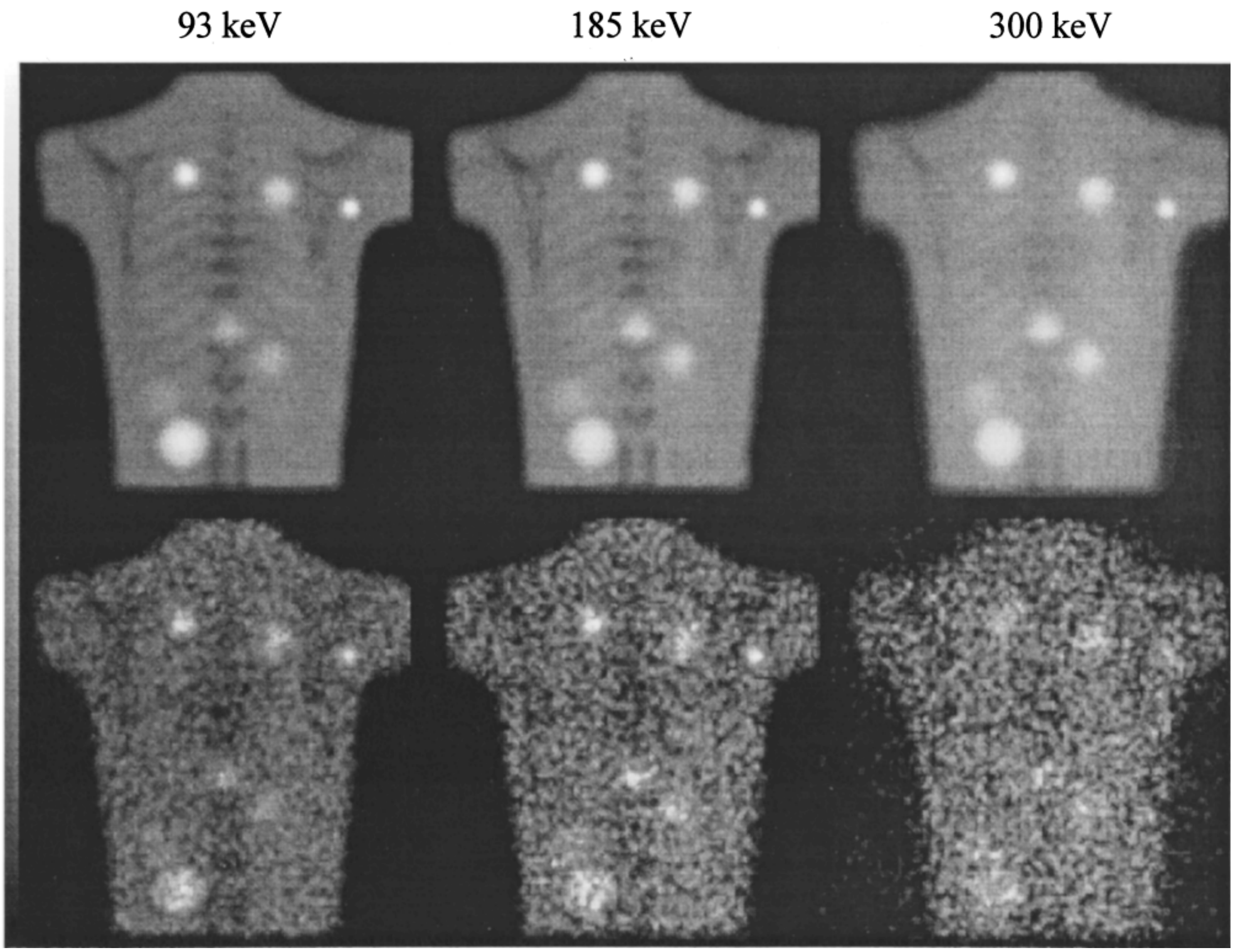

FIGURE 2. Noise-free, primary ~unscattered! distributions of the 93, 185, and 300 keV photopeaks ~60–370 keV! in an anterior projection of the torso phantom ~top row! and corresponding noisy projections ~bottom row!. All images are displayed using the same gray scale.