Published in final edited form as:

IEEE Trans Nucl Sci. 2009 October 6; 56(5): 2659–2671. doi:10.1109/TNS.2009.2028138.

C-SPECT - a Clinical Cardiac SPECT/Tct Platform: Design Concepts and Performance Potential

Wei Chang, Caesar E. Ordonez, Haoning Liang, Yusheng Li, and Jingai Liu

Department of Diagnostic Radiology, Rush University Medical Center, Chicago, IL 60612, USA

Abstract

Because of scarcity of photons emitted from the heart, clinical cardiac SPECT imaging is mainly limited by photon statistics. The sub-optimal detection efficiency of current SPECT systems not only limits the quality of clinical cardiac SPECT imaging but also makes more advanced potential applications difficult to be realized. We propose a high-performance system platform - C-SPECT, which has its sampling geometry optimized for detection of emitted photons in quality and quantity. The C-SPECT has a stationary C-shaped gantry that surrounds the left-front side of a patient's thorax. The stationary C-shaped collimator and detector systems in the gantry provide effective and efficient detection and sampling of photon emission. For cardiac imaging, the C-SPECT platform could achieve 2 to 4 times the system geometric efficiency of conventional SPECT systems at the same sampling resolution. This platform also includes an integrated transmission CT for attenuation correction. The ability of C-SPECT systems to perform sequential high-quality emission and transmission imaging could bring cost-effective high-performance to clinical imaging. In addition, a C-SPECT system could provide high detection efficiency to accommodate fast acquisition rate for gated and dynamic cardiac imaging. This paper describes the design concepts and performance potential of C-SPECT, and illustrates how these concepts can be implemented in a basic system.

Index Terms

Cardiovascular system; imaging; instrumentation; SPECT; system analysis and design

I. INTRODUCTION

Cardiac perfusion SPECT imaging currently accounts for about half of all clinical nuclear imaging performed in the United States. This is because coronary heart disease (CHD) is a common health problem and cardiac SPECT is often the first advanced imaging technique for evaluation of suspected CHD. However, the majority of clinical SPECT systems used for cardiac imaging today are still general-purpose systems that are not really optimized for cardiac SPECT. To achieve optimal cardiac imaging performance, a SPECT system needs to be designed for optimized imaging the heart, which is unique in its anatomy, location, and imaging requirements. Several new and dedicated cardiac SPECT systems that attempt to address these needs have emerged recently and are being promoted commercially [1], [2].

Because of the restrictions imposed by radiation burden and imaging time, nuclear imaging is inherently photon-limited. The primary performance-limiting factor of all SPECT systems is photon sensitivity or, specifically, detection efficiency. For cardiac SPECT imaging, desirable photons are scarce because typically only 1–3% of the injected radiotracer

localizes in the myocardium [3]. This fact largely explains the poor quality of cardiac SPECT images acquired in a standard 15-minute time frame. This low detection efficiency is even more critical for demanding imaging procedures, such as gated cardiac SPECT, which typically requires gating windows less than 0.1 sec. Nevertheless, the diagnostic utility of gated cardiac SPECT has been well established, and its clinical demand is still growing. A cardiac SPECT system with improved detection efficiency can be used to shorten imaging time to reduce incidences of motion artifacts and increase patient throughput, or allow reduction of radiotracer dosage and radiation exposure, while maintaining imaging quality and diagnostic accuracy. Furthermore, a much-anticipated imaging application that may open a new diagnostic dimension is dynamic cardiac SPECT. One example is imaging the heart in the first few minutes after injection of a fast-flowing tracer, such as Tc-99m-Teboroxime (Cardiotech) [4]–[8]. This procedure could allow quick and repeated assessment of perfusion, viability of myocardium, and cardiac reserve. Due to the low detection efficiency of current SPECT systems, this tracer's kinetics in human heart can only be studied with planar imaging but not with the SPECT technique yet. Due to rapid washout, the ultimate dynamic cardiac studies require high sampling rates, ideally with sub-10 second temporal resolution. For such fast acquisition, a largely stationary SPECT system with much increased detection efficiency is necessary. In initial explorations of clinical dynamic cardiac SPECT imaging, the current target is to achieve sub-2-minute acquisition using high detection efficiency systems [9]. Another approach that demands greatly increased detection efficiency to improve cardiac SPECT is using respiratory gating to reduce motion blurring [10].

For proper attenuation correction of SPECT images, a patient-and-slice-specific attenuation map derived from a transmission CT (TCT) is required [11]–[17]. Despite active development and commercial availability for years, TCT techniques have yet to be fully accepted by the clinical community. We believe the reasons are: marginal improvements, cumbersome operation, benefit-to-cost tradeoffs, and lack of adequate complementary scatter correction methods, etc. Nevertheless, two successive position statements published jointly by the American Society of Nuclear Cardiology and the Society of Nuclear Medicine strongly endorsed the use of attenuation correction and encouraged further development of TCT techniques [18], [19]. A cardiac SPECT system should have an effective and practical TCT component.

We have developed new design concepts for a cardiac SPECT/TCT system, which we now call C-SPECT (formerly called MarC-SPECT) [20]. These concepts have recently evolved into a second-generation design [21]. Thus, we consider C-SPECT as a new platform of SPECT systems. Our approach focuses on the system's geometry to optimize the detection of photons emitted from the heart in quality and quantity. The C-SPECT platform has potential to achieve much-improved performance with cost-effectiveness, offer simple and practical clinical operations and provide consistent and improved image quality. This platform also includes an integrated TCT component that shares the same detector system with SPECT operation. To distinguish it from the current hybrid SPECT/CT systems that combine two independent SPECT and CT systems, we use the label "Tct" to denote its secondary role in C-SPECT. The Tct is not of diagnostic quality on its own and is designed to provide attenuation maps for attenuation correction, to guide image co-registration, and to provide supports for image reconstruction.

II. GENERAL DESIGN CONCEPTS

The C-SPECT platform is centered on the following general design concepts.

A. Optimizing Detection Efficiency

Our basic assumption is that a SPECT system's performance potential is largely represented by its detection efficiency for a given sampling resolution. Detection efficiency is defined by the collimator system and is conventionally expressed as the detection solid angle for a point source. It can be viewed as a commodity that can be traded for sensitivity or spatial resolution with different collimators. We expand this concept to "system geometric efficiency (SGE)" as a characteristic parameter of a SPECT system. The subject of SGE is described in details in Section III-G. Our design principle is to maximize SGE under practical constraints. Because SGE is related to detector coverage, our concept is to provide as large a detector area as practical for detecting emission-photons.

B. Effective collimation and Sampling

To assure the quality of data, we implement effective collimation and sampling in an optimized system geometry. Basically, a large area detector in a body-conforming geometry is complemented with effective, while flexible, collimation and sampling of the heart region.

C. Predetermined Imaging Volumes

To realize the above two concepts, we have introduced a target-centric imaging concept through a new feature called predetermined imaging volume (PIV) [20]. Each PIV is a invisible cylindrical volumes defined in the patient aperture. The rest of the system is designed to optimize imaging the target in the PIVs. The details are described in Section III-D.

D. Adaptive Imaging

We use a brief scout-imaging first to select an appropriate PIV for each patient. SPECT imaging is performed after the heart of the patient is moved into the selected PIV. Thus, predetermined and optimized imaging parameters are selected adaptively to suit each patient and imaging situation.

E. Multiple Sequential Imaging

We introduce a collimator exchange scheme to facilitate multiple sequential imaging modes to be conducted efficiently without changing patient's imaging position.

F. Integrated Tct Imaging

A transmission-imaging scheme is implemented on the same detector system as a sequential imaging mode to SPECT imaging.

III. C-SPECT-I

From here on, we present a basic design of the C-SPECT platform, called C-SPECT-I, to illustrate potential implementations of these concepts. Because the detailed designs of C-SPECT-I have not been completely finalized, many design parameters are described with a range of values.

A. System Configuration and Gantry

A virtual C-SPECT-I system is shown in Fig. 1 with a patient being imaged in an upright position. It has a C-shaped stationary gantry, which houses the major components – the collimator and the detector systems. As shown in Fig. 2, in the transverse plane, the gantry wraps around the patient's left and front thorax and extends axially to cover at least the full length of the heart. The gantry's longitudinal axis and the patient's long axis are parallel to each other and slightly reclined. The front end of the gantry facing the patient is called the

C-arc. The open space partially enclosed by the C-arc is the patient aperture. The size and shape of the C-arc are designed with a general thoracic contour in mind to allow most patients to be positioned close to the collimator system. In C-SPECT-I, the shape of the C-arc, in the transverse plane, is based on a modified partial ellipse.

B. Collimator System

A collimator system resides at the front of the gantry conforming to the curvature of the C-arc. The collimator system defines the imaging field and is designed specifically for a PIV. The transverse plane of the collimator system and its associated PIV are shown in Fig. 2. The collimator system is based on the “slit-slat” design first introduced in the Michigan’s brain SPECT system SPRINT- II [22] – [24].

1) Slit-Slat Collimator—The collimator system consists of two physically separate components - a slit-plate and a slat-stack - shown separately in Fig. 3. The slit-plate is basically a curved lead plate (6 to 8 mm thick) with multiple axially oriented slits, which are air gaps of uniform width. The slat-stack is a stack of thin lead plates or foils (about 0.2 mm thick) parallel to the transverse plane, and separated by about 3–4 mm of air-equivalent gaps. The slat-stack extends radially, with varying length from 8 to 10 cm, and fills most of the transverse space between the slit-plate and the detector system. The axial length of the slat-stack is 16 cm. Each slit casts a minified non-overlapping projection of the PIV on the detector surface. These projections are inverse-fan-beam collimated transversely and parallel collimated axially. In the transverse plane, the collimator system subtends an angle of approximately 200°, from the first to the last slit, with respect to the center of the primary PIV - the frequently used PIV.

2) New Features—We have modified the original slit-slat collimator to suit C-SPECT-I as follows:

- a. *Stationary during imaging:* Keeping the slit-plate stationary during SPECT imaging is desirable to maintain a well-defined PIV while optimizing the use of detector area. This is because this slit-plate, which conforms to the curvature of the C-arc, is not circular; a small step-and-shoot rotation, as operated in SPRINT-II, would cause successive projections of a slit to shift away from the intended PIV. This stationary feature facilitates precision collimation, robust calibration of the collimator-detector system, and is not vulnerable to the common center of rotation problem in rotating-camera SPECT systems.
- b. *Number of slits:* The number of slits, and their corresponding number of projections allowed on the detector-arc, is determined by the projection’s width. It is in turn related to the projection minifying factor and the intrinsic spatial resolution (ISR) of the detector system. For ISR in the range of 2.0 to 3.5 mm, the number of slits in C-SPECT-I is in the range of 10 to 16.
- c. *Slit width:* The transverse collimator resolution of each slit is determined by its width. The widths of the slits are about 3 mm. A desirable feature is that each slit’s width can be adjusted individually to yield the same transverse collimator resolution, or sampling resolution, at the center of its associated PIV.
- d. *Slit-guides:* Both sides of each slit on the slit-plate are knife-edge-shaped. In addition, a pair of divergent lead pieces, called slit-guides, will be added to both sides of each slit, as shown in Fig. 3a. These slit-guides restrict incoming photons to those from the PIV and prevent overlapping of adjacent projections on the detector. The orientations of the slit-guides, varying from slit to slit, define the size

and location of its associated PIV. On the transverse plane, the two slit-guides of each slit are symmetric to a line that connects the slit to the center of its PIV.

C. Detector System

The detector system is located immediately behind the slit-stack of the collimator system, as also shown in Fig. 2. The detector system is also generally C-shaped and is assembled from a series of detector modules. Room-temperature semiconductor detectors, such as CZT-ASIC [25]–[28], or pixellated-scintillator coupled to PSPMT or PD/APD [29]–[31] can be readily used here, especially in view of their high intrinsic spatial resolution (ISR) and lack of edge effects. However, they are currently far too expensive for the large detector area required here. To be cost-effective, our detector approach is based on conventional technology. The detector system of C-SPECT-I will be assembled from many (10–20) NaI(Tl)-based detector modules, packed side-by-side transversely. Each module is covered by an array of medium-size single-anode PMTs. The total angular coverage of the detector-arc is slightly larger than that of the slit-plate, about 210° with respect to the long axis of the primary PIV.

The detailed design of the detector module has yet to be finalized. It is beyond the scope of this paper. Nevertheless, the general requirements of the detector system are clear and presented as follows.

1. *Large Detector Area:* To achieve large detection solid angle, and hence high SGE, the total detector area will be as large as practical, under constraints of cost, and functional and operational concerns.
2. *High Packing Fraction and Stopping Power:* To catch emission photons efficiently and effectively, the large-area detector system should be practically continuously position-sensitive with minimal detection gaps. Appropriate detector thickness is essential to provide adequate stopping power for the photon energies of interest. For our major application with 140 keV photons, the thickness of the NaI(Tl) can be 10 mm. It is important to eliminate, or at least minimize, the edge effects on the transverse plane of the detector modules to minimize sampling gaps and maximize detector area.
3. *High Transverse Intrinsic Spatial Resolution (ISR):* The high SGE of C-SPECT results from using a large number of minified projections of a relatively small PIV. The prerequisite for using minified projections is that the detector has adequately high ISR to sample the high spatial frequency information of the minified projections. Because minified projection is used only in the transverse direction, the detector's transverse ISR requirement is demanding. To achieve high ISR (< 3 mm) with 10 mm thick NaI(Tl) slab, appropriate segmentation or pixellation of the slab is necessary.
4. *Decent Energy Resolution:* Good energy resolution of the detector system reflects the effectiveness of a detector system's light collection and utilization as well as facilitates energy discrimination. Our targeted energy resolution at 140 keV will be in the 9–11% range, comparable to other NaI(Tl) based segmented detector systems, such as Anger cameras, SPRINT-II, and current pixellated detectors.
5. *High Count-Rate performance:* Due to the modular nature of the detector system, each detector module operates independently and in parallel with other modules. This architecture allows the detector system to achieve high CR capability even with standard electronics used in Anger cameras. For very high CR performance, special electronics, such as HYPER architecture, can be implemented [32].

D. Imaging Operation

1) Predetermined Imaging Volume (PIV)—All the operation is centered on imaging a target located in a specific predetermined image volume (PIV). This concept is extended from our previously proposed cardiac SPECT system [33]. Due to variations among patients, several PIVs are needed to accommodate a general patient population. The sizes and locations of these PIVs, should be based on a survey of cardiac and thoracic anatomy of cardiac patients under the same imaging position. For example, the PIV shown in Fig. 2 is a 15-cm diameter cylindrical volume, which is designated as the primary PIV that suits the majority of patients. In contrast, Fig. 4 shows this primary PIV along with a larger 18-cm diameter PIV, which is shifted slightly to the right to accommodate a larger patient and/or a larger heart. For two other imaging modes (scout-imaging and Tct) to be discussed later, two much large PIVs will be defined, as scout-PIV and Tct-PIV, respectively by their associated slit-plates.

2) Patient Rotation for Sampling—In many SPECT imaging situations with C-SPECT-I, the required number of projections exceeds a single set of projections of a slit-plate. Thus, rotation of the patient with respect to the PIV's axis can provide additional projection sets to meet imaging requirements.

3) Scout-Imaging—To position a patient's heart into a PIV for SPECT, a scout-imaging without patient rotation is performed first at the initial default position. It requires a special slit-plate, which has a smaller number (5–7) of slits to sample a large PIV (of 32–44 cm), as shown in Fig. 5. The slits are much wider for high efficiency/low resolution imaging. After adequate photon statistics is acquired in 1–2 min., the center coordinates and the size of the heart can be estimated in real-time from the reconstructed scout-images.

4) Smart-Chair System and Automated Patient Positioning—To facilitate 3D translation and rotation of patients, a computer-controlled Smart-chair system plays the key role. The chair will have 6 degrees of freedom and is mounted, through a 3D translation stage, on a heavy-duty turntable whose longitudinal axis is perpendicular to the transverse plane of the gantry. After scout-imaging, the Smart-chair system will select an appropriate PIV and move the patient's heart into it. As the patient's heart is moving onto the turntable's axis, the axis of the turntable also shifts laterally to align with the axis of the selected PIV. Therefore, when patients are rotated on this axis in the following SPECT imaging acquisition, their heart keeps remaining in the PIV. After SPECT, the subsequent Tct employs a different and much large PIV. Because the sizes and locations of the two PIVs are different, the Smart-chair needs additional two degrees of freedom to move the patient between the two axes of their respective PIVs.

The Smart-chair should come with ergonomically designed armrests, firm and comfortable back support and motion restraining devices. The gantry is tapered radially toward the patient on the longitudinal plane, as shown in Fig. 1, for patient's arms to raise and rest on the elevated armrests.

5) Data sampling and reconstruction—The non-circular contour of the C-arc and the variable distance between the slits to the PIVs lead to two unconventional variations in data sampling. The first is uneven angular sampling, because the slits are not evenly spaced angularly with respect to the center of all the PIVs. The second is unequal fan-angles among the sampling-fans formed by different slits. These two variations imply that iterative algorithms, such as OSEM, rather than conventional FBP, are necessary for image reconstruction [34]. In addition, iterative algorithms can handle quite effectively the

truncated projections caused by sampling a small and isolated volume in the patient's body [35]–[38].

E. Fast and Automated Collimator Exchange

The C-SPECT platform is designed to perform three different imaging modes (scout-imaging, SPECT, and Tct) in sequential phases. Each mode requires a specific slit-plate to define a PIV and to meet its collimation and sampling requirements. Thus, we need a scheme to change slit-plates, quickly and smoothly, without disturbing the patient. To facilitate this operation, the slit-plate can be made semi-flexible. For example, the slit-plate can be constructed as a chain of narrow segments of 2–3 cm width in the transverse plane. Each segment overlaps with its neighbor segments to prevent radiation leakage. An analogy would be that of household vertical Venetian blinds (V-blinds) in closed position; each vertical-slat (or v-slat in V-blinds terminology) of the blinds is partly in contact with its adjacent v-slats. This slit-plate chain-assembly is mounted on rollers that slide on tracks fixed to the top and bottom frames of the gantry. Moreover, multiple slit-plate-chains can be connected in series on a conveyor belt and looped around the detector system, as also shown in Fig. 5. A computer-controlled stepper motor can turn the conveyor belt with precision. Care must be taken in the design of the conveyor belt to achieve accurate and reproducible collimation for the slits. The gantry can be designed to accommodate four slit-plates in a loop; e.g., one for scout imaging, two for PIVs of different sizes or locations, and one for Tct imaging. Provision can be made for a second track, connected as a branch of the main track for additional slit-plates. A control system can select a specific slit-plate on a track to move up to stage for imaging.

F. Integrated Transmission CT (Tct) System

Integrated in the C-SPECT-I is a Tct system. The Tct system uses the same detector system for imaging photons transmitted from a source on the other side of the patient, as shown in Fig. 6. The Tct can be performed with a moderate acquisition time of 1–2 min. to keep patient comfortable and to average out patient's respiratory motion to better match its images with the separately acquired SPECT images.

1) Sampling options—Fan-beam Tct typically requires an angular sampling range of 180° plus the fan angle. Assuming a typical fan-angle of 50° , a total of 230° rotation of the source relative to the target is needed. The primary scheme is to rotate the patient 230° , as shown in Fig. 6. Tolerance for this large and slow rotation may not be a problem for most patients, provided they are comfortably and firmly secured in the chair. As an alternative option, the rotation can be shared between the patient and the transmission source; both rotate synchronously on the same axis but in opposite directions. This option will be explored if significant number of patients prefer, or can only tolerate, a smaller angle (such as 100° – 120°) of rotation. Its implementation needs to be weighted by increased mechanical complexity and costs.

2) Slit-plate for Tct: the V-blinds—When conducting Tct after SPECT, the slit-stack remains in place, but the slit-plate used for SPECT must be moved out of the way to allow transmitted photons to reach the detector. A special slit-plate will move into position for Tct imaging. This slit-plate would look like an “open” household Venetian-blinds, again shown in Fig. 6. It is made up of a series of open v-slats, which hang in front of the slit-stack for transverse collimation. These v-slats are thin lead foils separated by air-equivalent material and can be 10 – 15 mm deep and spaced about 2–3 mm apart. On the transverse plane, all the v-slats focus to a longitudinal transmission line source in a fan-beam. The V-blinds not only block scattered photons and allow unscattered photons in the fan-beam to go through.

More importantly, they reduce emission photons from patient's body from swarming the detector system.

For the first Tct option that requires patient rotation only, the V-blinds set remains stationary throughout imaging. For the second Tct option with combined source and patient rotation, the V-blinds set would rotate synchronously with the rotating transmission source to maintain their focus on the source. Since these V-blinds set can also be assembled as a semi-flexible chain, the v-slats can be grouped into segments that are also mounted on rollers and slide on the same tracks. Because each v-slat is rigidly fixed on its roller, its orientation is solely defined by the local track and remains largely focused to the source as both rotate over a limited angle, such as $\pm 60^\circ$, from its middle position.

3) Tct source—To take advantage of the high-count rate capability of the detector system, we use a high-intensity source for Tct. A high intensity scanning point source (for longitudinal scan at each rotational step) or a stationary line source can be used. Our choice is to use an x-ray tube to provide filtered x-rays or characteristic x-rays when a heavy-element is irradiated [39], [40]. In both cases, the transmission flux is reduced in intensity and centers on a narrow energy band suitable for photon counting detection.

G. Performance Potential

To evaluate the performance potential of the C-SPECT platform, we estimate its system geometric efficiency (SGE) and verify the adequacy of sampling schemes for artifact-free imaging. We use calculations and simulations based on a tentative model of C-SPECT-I. A relative SGE, or SGE index (SGEI), can be derived by comparing with that of a benchmark – a model of a dual-head SPECT system.

1) System geometric efficiency (SGE)—The geometric efficiency (GE) of a collimator is defined conventionally as the number of photons passing through the collimator as the fraction of the photons emitted from a point source [41], [42]. The SGE of a SPECT system is the sum of the GEs from all its collimator components with respect to an appropriately located point source. Basically it is the total detection solid angle subtended at the point source when it is sampled, by the detector and collimator systems, at a specific spatial resolution. Thus, the SGE of a system has a well-defined value that can be calculated directly from the system's model and parameters, including collimator design and target distances explicitly defined in the model.

Consider a slit-slat collimator with a single slit, such as that shown in Fig. 3. Its performance metrics are expressed in newly published formulas [24], [43]. Denote by d the width of the slit; by t and L , the thickness and length of each slat, respectively; and by h the separation gap between adjacent slats. Consider a point source at a perpendicular distance b from the slit-plate, and a perpendicular distance b' from the front of slats, and at an angle θ with respect to the normal to the slit-plate. The on-axis transverse and longitudinal collimation resolutions are given by

$$R_{ct} = d_e(1+m) \quad (1)$$

$$R_{cl} = \frac{h_e(L_e + b')}{L_e} \quad (2)$$

respectively, where

$$m = \frac{b}{L_t} \quad (3)$$

and is called the minifying factor. The subscript e in some parameters above denotes the use of corrected values after taking into account the effects of septal penetration. In each direction, the sampling resolution (with subscript s) is given by

$$R_{sx} = \sqrt{R_{cx}^2 + (mR_{tx})^2} \quad (4)$$

where R_{tx} is the intrinsic spatial resolution along that direction (subscript: $x=t$ for transverse, and $x=l$ for longitudinal). In the longitudinal direction, $m=1$, the first factor is much larger than the second factor; the sampling resolution is thus mainly determined by the collimation resolution (with subscript c). In the transverse plane, $m > 1$, the second term becomes significant. To keep the sampling resolution under control in the design of C-SPECT-I, we need to select m appropriately to go with the transverse intrinsic spatial resolution, R_{tb} to contain the second term. The smaller (higher resolution) the R_{tb} , the larger the m , and more number of slits and projections, can be used. Finally, the geometric efficiency of this single-slit slit-slat collimator is given as

$$GE = \frac{d_e h_e^2}{4\pi b L_t (h_e + t)} \cos^3 \theta \quad (5)$$

a) Conditions of Calculation: When calculating SGE, the imaging condition of the system is directly relevant and should be stated explicitly [20], [44]. For comparison of SGE, ideally, systems should meet the conditions of achieving the same imaging performance. In practice, we use a simple condition for all systems: sampling a central point source with the same reference spatial resolution from all directions.

i) Sampling spatial resolution: Because it is not possible to make different systems to perform under exactly the same sampling resolution, we use modeling to normalize their sampling resolution virtually on paper. In the following calculations, we fix the 3D sampling resolution of a point source at the center of the heart to be 11 mm FWHM isotropically. For the dual-head SPECT model, sampling resolution of the point source cannot be isotropic due to variation of sampling distance in different projections, especially with a body-contour orbit. Therefore, we approximate this condition by requiring the “average sampling resolution”, from all directions, of the point source to the same 11 mm FWHM.

ii) Location of point source and the thorax model: In calculating SGE, the point source is placed at a proper location in the SPECT system, such as where the center of the heart would be during imaging. We define this location using a thorax model, represented by a 40×30 cm ellipse, where the center of the heart is located at 6 cm to the left-anterior 45° direction from the center of the ellipse.

b) C-SPECT-I model: We used a model for C-SPECT-I with 12 slits in this calculation. The patient is positioned so that the minimum clearance is 3 cm. The detector–arc is assumed to be continuously position-sensitive with no functional gap. The slits are arranged such that, in the transverse plane, they sample data simultaneously of a PIV with non-overlapping projections. To achieve a uniform transverse resolution at the center of the PIV, the minifying factor for all the slits is set to the same fixed value, 1.85.

Next we calculate the SGE for a point source at the center of a 15-cm diameter PIV, as shown in Fig. 2. Table I lists the imaging conditions in the model and basic parameters of the slit-slat collimator. Table II summarizes the other relevant parameters and calculated results for the odd-numbered slits. The first row in Table II shows the radial distances of the odd-numbered slits from the center of the PIV. The fourth row shows the calculated GE for each slit based on Eq. 5. The last row shows the sum of the GEs from all 12 slits. Thus, our C-SPECT-I model predicts an SGE of 56.4×10^{-5} .

c) Dual-Head SPECT model: For a dual-head SPECT system, we use a model shown in Fig. 7. Two gamma cameras in L configuration, each represented by its parallel-hole collimator, rotate over 90° in 32 steps around the patient from the RAO-LAO projections. The patient model is the same 40 cm x 30 cm thorax. A special feature is the contour orbit that the cameras follow; it is an elliptical arc that clears the thorax by 3 cm. At each step, the target distances were measured, and the sampling resolution was calculated accordingly. Because the sampling resolution at the center of the heart is not uniform among the projections, we adjusted the collimator parameters of this dual-head model iteratively until the average sampling resolution from all 64 projections to the point source reach 11 mm FWHM. Table III lists the sampling resolution of 8 equal-spaced steps of the 64. With the collimator parameters defined, the distance-independent SGE ($=28.2 \times 10^{-5}$) of the two identical parallel-hole collimators is derived from the Anger formulas [41], [42].

d) Comparison between the two models: Table IV compares the SGEs of a typical dual-head SPECT model with three C-SPECT-I models, each having a progressively improved detector ISR to accommodate 11 to 13 slits. These C-SPECT-I models yield SGEI from 1.91 to 2.05.

2) Simulated Imaging—The other major concern in the system design is the adequacy of data sampling. Hence, sampling and image reconstruction of C-SPECT-I were investigated through simulations. The areas examined are number of rotational steps, sampling pattern, total angular sampling range, and their effects on reconstructed image quality.

a) A digital heart-body phantom: The simulations used a digital phantom, shown in Fig. 8a (in the conventional feet to head orientation), with its magnified central portion shown in Fig. 8b. The target organ is the “myocardium,” represented by a 10-mm thick, 120-mm diameter annulus with uniform concentration of radioactivity. The annulus encloses 12 circular hot-spots of the same activity concentration but various sizes (6, 8, 10, and 12 mm diameters), separated from its neighbor by the sum of the diameters of the two adjacent hot-spots. The center of the 256×256 matrix in Fig. 8a is the center of the annulus, the center of the middle 10 mm hot-spot, as well as the center of the 150 mm diameter PIV (not shown). The patient body, represented by a circular 250-mm diameter disk, has its center offset 5 cm to the 45° RPO direction from the center of the annulus. The body contains a uniform background of 20% concentration of that in the annulus. To the right of the myocardium is a small elliptical disk, representing the dome of a liver, at 60% concentration of that in the annulus.

b) Simulations for C-SPECT-I: The sampling resolution from all directions toward the PIV is assumed to be a constant 11 mm FWHM with no distance dependence. The simulations also assumed ideal data collection with no other degrading effect caused by attenuation, scatter, noise, and filtering or other processing. The following aspects are simulated and evaluated.

i) Number of rotations: The required number of angular sampling in CT is proportional to the size of the FOV centered at the axis of rotation and the maximum spatial frequency to be imaged [45]. C-SPECT takes advantage of this relationship. With a relatively small PIV located on the axis of rotation, it requires a smaller number of angular sampling than conventional SPECT systems to provide the same imaging resolution. Our following simulation is based on a C-SPECT-I model with 195° angular coverage between the two end-slits to the center of the PIV. Ordered-subset expectation-maximization (OSEM) algorithm was used to reconstruct the 20×20 cm region of the phantom centered at the PIV [34]. The reconstruction also used an *a priori* body-contour as a support, which assumed no activity outside. In Fig. 8c to 8f, the 4 images are reconstructed with numbers of angular sampling from 12, 24, 36, to 48, corresponding to 0, 1, 2 and 3 rotations of the phantom, respectively. The corresponding angular sampling ranges are 195° plus the rotation angles, which are 0° , $1 \times 10^\circ$, $2 \times 7^\circ$ and $3 \times 5^\circ$, respectively. As expected, image quality improves with the number of angular sampling. The image in Fig. 8c, for the case of no rotation (12 projections), shows poor quality. Thus, stationary imaging operation in C-SPECT-I does not provide quality image under this simulated sampling resolution (of 11 mm) without further processing. Rotation of the target can increase angular sampling to improve image quality. The images, in Fig. 8e and 8f, with 3–4 sets of data (36–48 projections), show excellent and similar image quality.

ii) Range of angular sampling: In theory, complete sampling of a PIV in the SPECT mode requires an angular coverage of 235° (i.e., 180° plus a fan angle, which varies between 40° – 55°) with respect to the center of a PIV. However, such a large angular coverage in C-SPECT-I would make the gantry too restricted for proper patient positioning and maneuvering. The appropriate range is an important issue that directly relates to the physical size of the gantry. In the simulations, these angular ranges are defined with respect to the center of the primary PIV and subtended symmetrically to the 45° LAO direction of a properly positioned heart-body phantom. Imaging simulations were conducted for 5 angular-sampling ranges: 180° , 195° , 210° , 225° and 240° . Note that only the last one meets, and actually exceeds, the complete angular sampling requirement. In each case, four sets (48 projections) of projection data, simulated for the scheme involving three 5° rotations, were reconstructed. Three of the reconstructed images (for 180° , 210° , and 240°) are shown in Fig. 9. They demonstrate adequate image quality for all these sampling ranges. The 180° image, in Fig. 9a, shows slightly decreased activity and minor artifacts in the far backside of the annulus. This image can still be considered marginally acceptable, even with less quality, because the backside of the PIV has no radiotracer uptake in clinical situations. The other images, with angular-sampling ranges equal to or larger than 195° , are artifact-free and visually identical; they also yield nearly identical quantitative profiles inside the PIV, as shown in Fig. 10. Visual differences, as well as difference in profiles, are mainly outside the PIV in the rest of the body, and are the results of the grossly incomplete sampling and severe projection truncation in those regions.

Fig. 11 shows a semi-quantitative assessment of the images acquired with the different angular-sampling ranges. It plots the square root of the sum-of-squared-error inside the PIV, between the 5 reconstructed images and the digital phantom, as a function of angular-sampling range. The largest error occurs at 180° . Increasing sampling range beyond 180° leads to small errors and are quite comparable to that of 240° , which has complete sampling. The remaining small differences are attributed to round-off errors. We conclude that the angular-sampling range of 195° is the minimum required for C-SPECT-I.

iii) Sampling pattern in Sinogram space: The sampling pattern in Sinogram space with 3 sets of projection data (two 7° rotations) for the 12-slit C-SPECT-I model is shown in Fig. 12. The detector-arc is assumed to be sampled every 3 mm along its transverse plane. The

abscissa is the radial positions of the projections from each slit relative to its own central axis, which is a line connecting the slit to the center of the PIV on the transverse plane. The ordinate shows the corresponding projection angles, in degrees, measured from a reference axis. There are twelve distinct slanted bands, one for each slit, in the pattern. Each of these bands consists of 3 slanted lines, one for each projection at a rotational step. The figure shows that the sampling density in the sinogram space, although not exactly uniform, is quite different from, and more uniform than, that of the SPRINT-II [22]. The total angular sampling range of this case is 214° , which is the sum of 200° between the two outermost slits to the center of the primary PIV and 14° of the 2 additional rotations. There are several local blank spaces in this sampling pattern representing missed data points, which are the result of physical gaps between adjacent detector modules in the detector-arc. In this model, we assumed a 3 mm sampling gap between adjacent modules. Each of the detector modules is assumed to be 10 cm wide and its scintillator slab is fully position-sensitive in the transverse plane of the module. Because the width of the module does not match with the projection width, the blank spaces are located at different relative positions in different projections. Although there are also several thin over-sampled, but not exactly double-sampled, slanted stripes, all these data are independent measurements and can be used effectively in reconstruction and contribute to image quality.

IV. DISCUSSION

In the C-SPECT platform, we start out from a basic system geometry that detects photons, efficiently and effectively, from a number of predetermined imaging volumes (PIVs). Effective sampling for imaging is implemented through efficient and flexible collimation. A central concept is the multiple PIVs and their appropriate and flexible placement. The size and location of the primary PIV should be adequate for most of patients' hearts [46]. Additional PIVs with different sizes and locations can be defined for other segments of patient population. It is this flexibility – selection of one of the PIVs using a scout-imaging – that makes this rigid system geometry practical and optimized for imaging a general patient population.

Another important design parameter is the appropriate range of angular sampling, which determines the angular span of the gantry and the C-arc. The latter should be sufficiently large to image most patients properly, while small enough for cost and operational concerns. Our simulations indicate that an angular-span of $195^\circ - 210^\circ$ between the two end-slits is appropriate. For stationary imaging with no target rotation, the C-SPECT-I will provide at least 200° of angular sampling of the primary PIV. With target rotation, an additional sampling angle of $10^\circ - 15^\circ$ would be added to yield $210^\circ - 215^\circ$ of total angular sampling.

For typical clinical imaging, when many degrading factors are present and resolution requirement is lower than we have used in our simulations, the total number of projection sets could be reduced to 2, i.e., with one rotation only. For dynamic imaging with fast framing rate (at 5 or 10 sec), a C-SPECT system can operate in a completely stationary mode, or in a one-rotation mode alternating between two target orientations, based on image quality requirements. The raw data contain valuable timing information for post-processing applications, such as 4D compartmental modeling or deriving SPECT images of flexible temporal and spatial resolution. In data sampling, the feature of a small number of target rotations leads to many practical benefits. Not only the idling time during rotation is minimized, it also makes the data set more compact in size (such as $2 \times 12 \times 32 < 0.8K$, when using 2 projection sets with 1 rotation), which is smaller than that of all other contemporary approaches by a factor between 2 to 15 in our estimation. This is a desirable feature in reconstruction time, especially for iterative algorithms. More significantly, this compact data size implies the data are made up of larger and more accurate numbers with reduced relative

statistical noises, because the total counts are distributed over a smaller number of pixels. This low noise feature boosts the signal to noise ratio, and when compounded with the low noise resulted from increased SGE, lead to high quality images in C-SPECT. This low noise feature may allow acquisition frame rates to be increased further. More investigations are needed to investigate this possibility and its limitations.

In C-SPECT-I, the point-spread-function at the center of the PIV will be quite isotropic. Because of the stationary gantry and rotation on the axis of the PIV, target distances remain unchanged in imaging. Thus, sampling resolution at the center of the PIV from all directions can be matched by adjusting the widths of the slits. As a result, the 3D image resolution throughout the PIV will be quite uniform. This relatively uniform spatial resolution is very desirable, in light of artifacts and difficulty in interpretation caused by nonuniform and orbit-dependent resolution of conventional rotating SPECT systems [47]–[49].

The upright gantry and patient's sitting imaging position are preferred in C-SPECT-I. These features facilitate patient translation and rotation in selecting and imaging in a PIV, and are especially suitable for the large-angle patient rotation needed in Tct imaging. Although it is not impossible to develop a horizontally oriented system geometry that rotates the detector and/or slit-plate while keeping patient supine and stationary, the current upright design of C-SPECT-I is mainly due to cost issues and ease of implementation. Furthermore, supported by several preliminary studies, upright cardiac SPECT imaging may have the advantage of lowering the patient's diaphragm and breasts to reduce the severity of attenuation and scatter contamination caused by sub-diaphragm tissues and tracer distributions [50]–[52].

The C-SPECT-I system requires the development of a mechanically complex Smart-chair system to execute its imaging operation. The Smart-chair's functionality is the main reason that the designs of other components can be simplified, cost reduced, and performance improved. The required motions of the Smart-chair, although complex, are not more complex than those already existing in conventional high-end dual-head SPECT systems. These latter systems require a combination of typically 4–5 simultaneous motions during imaging, including detector rotation, lateral gantry and/or vertical couch translation, and two independent radial translations of detector heads for patient-specific contour orbiting. For systems with simultaneous SPECT/Tct, additional motion of transmission source(s), with sliding-window energy-selective acquisition, are involved during imaging as well. Because of the heavy load of these systems (1000–2000 lbs for detector system), the required precision motions and necessary patient-safety measures become complex and expensive. In SPECT systems, mechanical complexity is necessarily employed to expand functionality and reduce costs. In C-SPECT-I, the mechanical complexity is mainly confined to the lightweight Smart-chair system. We assume that patients can hold their thorax still during imaging on the Smart-chair. Thus, considerable effort must be devoted to designing the Smart-chair for patient comfort and immobilization.

In addition to using a point source model in estimating detection efficiency, we have used a more realistic 2D disk-source model with attenuation in comparing the SGE with the dual-head model. The latter model yields 5% less SGE than that of the point-source model for C-SPECT-I. This is because off-axis photons reach the slits obliquely, encountering an effectively smaller slit, to encounter the $\cos^3 \theta$ effect. Because the θ in the C-SPECT-I model averages only about 10° , its effect is relatively minor.

The concept of using minified projections is to allow more simultaneous projections and thus increase SGE. It was first implemented in 1D in SPRINT-II, and extended to 2D by the Arizona group [53]. Because a detector performance of 3.0 – 3.3 mm ISR is not difficult to achieve, a C-SPECT-I with 11–12 slits is realistic. Further increasing the number of slits and

minifying the projections can push the SGE higher, albeit only slightly due to other design constraints. In general, the SGE of C-SPECT-I doubles that of the conventional dual-head systems for cardiac SPECT imaging. Thus, a C-SPECT-I should at least provide the same imaging quality in half the time of a dual-head SPECT system in cardiac imaging.

Slit-slat collimation has been around for years, but has largely been overlooked because its niche is limited. Our C-SPECT-I exploits this collimation's strength and flexibility. Pinhole collimation can also be implemented in C-SPECT, in light of the recent success of small animal SPECT systems. In fact, pinhole-based clinical SPECT systems have received renewed interests recently [54], [55]. However, since projection magnification is not realistic for clinical SPECT systems, using pinhole collimation for C-SPECT requires 2D minified projections for adequate sampling. However, to achieve a high SGE and adequate 3D sampling, a large number of pinholes and a large detector of very high 2D ISR (probably < 2 mm) will be needed to accommodate the large number of minified projections simultaneously.

Our proposed collimator system and collimator-exchange mechanism make C-SPECT a versatile and powerful system by opening up imaging tasks and protocols that were not possible or not practical so far. With this exchange scheme, multi-phase imaging can be performed quickly and smoothly without disturbing the patient. As a result, images acquired in sequential phases can be matched anatomically and collimation can be selected to suit imaging requirements and adapt to patient variations. This concept is the same as the "Adaptive SPECT" proposed by Barrett *et al* recently [56]. In scout-imaging mode, a set of low-resolution/high-sensitivity images is acquired first in a large FOV. Although the purpose of scout imaging is to facilitate patient positioning, these low-resolution images are rough estimates of outside activity distribution that can help improve SPECT images. These images can guide corrections of errors introduced by truncation and scatter of SPECT images, when used in combination of Tct images [57].

Further increase of SGE in C-SPECT can be achieved by extending to 3D sampling. An example is replacing the parallel slats with longitudinal-converging slats. The resulting sampling is a bi-focal cone-beam geometry. This geometry can increase SGE by another factor of 2, depending on the convergence rate [58]–[62]. This feature will be explored in our second-generation design, C-SPECT-II [21]. Several cone-beam collimation approaches support the contention that increased 3D geometric efficiency leads to improved lesion-detection accuracy [63], [64]. Although cone-beam collimation has been investigated for years, it has not been implemented on clinical SPECT systems yet. The major difficulty is in keeping the heart in the cone-beam when detectors rotate. This is not a problem for our proposed C-SPECT-II, because the heart is locked in a PIV throughout imaging, even with patient rotation is involved. Thus, the potential and benefits of cone-beam imaging should be more likely realized in a C-SPECT-II system.

Our proposed Tct scheme also promises high performance-to-cost ratio. The low cost comes from the feature that SPECT and Tct modes share the same detector system. High performance comes from low-noise Tct images that result from the combined use of a high-intensity transmission source and a high count-rate modular detector system. The use of a high-intensity source, albeit at low energy, for Tct has several advantages. High intensity transmission can overshadow down-scattered emission photons, and reduce the scatter contamination to a small and insignificant fraction. If necessary and desirable, cross-talk contamination in the low-energy window can be corrected quite effectively [65], [66]. Using filtered or characteristic x-rays as the transmission source is our choice, which offers many practical advantages. These advantages include no need for regular replacement of the source, easy storage and handling, reduced concerns for radiation leakage, monitoring and

disposal, adjustable intensity to suit patient size and detector performance, and last but not the least, high reliability and low cost. Our modular detector system, when operating in a high-speed photon counting mode, can handle filtered or characteristic x-rays, which would be far more intense than any radionuclide-based transmission sources can provide.

In the last few years, several dedicated cardiac SPECT systems with new system geometries have been introduced and promoted commercially [67]–[69]. Our C-SPECT-I looks quite similar to the system – CardiArc; both systems are extensions of Michigan’s SPRINT-II. However, the similarities exist only on the surface, specifically, in the general system geometry and basic collimation. The systems are very different in all other design concepts and implementation, including the specific system geometry, the detector system, the collimator design and operation, the image volume, and imaging operations. The most significant difference is in their sampling operations. CardiArc uses detector-based sampling, similar to that of 4th-generation CT, while C-SPECT uses projection-based sampling, similar to that of 3rd-generation CT. Thus, these two approaches have quite different maximum framing rate, which is an important consideration in dynamic imaging. Based on our modeling studies of the generic sampling geometries of these new cardiac systems, their SGEs all turn out to be comparable (i.e., within 10 %) to the conventional dual-head model [44]. These studies were based on similar modeling techniques described in Section III-G. These findings are not unexpected, because the SGEs of SPECT systems are basically proportional to their detector area and inversely proportional to the size of the imaging field, when sampling resolution is normalized and the secondary differences of target distances are ignored. This assertion is supported by the well-established Space-Bandwidth-Product concept [70], [71]. In other words, with fixed bandwidth (i.e., resolution), the detection solid angle or detector area determines the quantity of information gathered. In SPECT systems, there is no substitute for detector area for providing detection solid angle. The discrepancies of system sensitivities between the relative SGEs of our modeling study and commercial claims are the results that the sampling resolutions were not normalized. One of the systems uses high-sensitivity collimators to sample low-resolution data, which are then reconstructed through a proprietary algorithm with a heavy resolution-recovery component [69].

With regard to resolution-recovery processing of SPECT images, it is a legitimate and powerful technique when used properly. Major commercial vendors have already embraced these techniques and have been promoting them in their products. They claim that these techniques effectively allow image acquisition time to be cut in half, with the same collimators, to achieve the same image quality. Although software processing is a crucial component that needs to be optimized for all SPECT systems, evaluation of software techniques has not been standardized. As a result, software processing is hard to assess rigorously and comprehensively. If these techniques can handle high sensitivity/low-resolution data effectively, it would be an exciting new direction for system design. We believe the basic principles of most software-processing techniques are not platform-specific. To compare different systems fairly and objectively, the performance of hardware and software should be assessed separately, although this condition is necessary but still not sufficient. The ultimate evaluation should include realistic task-based evaluation. Software processing for short acquisition data to gain equivalent diagnostic information is certainly welcome by all. Based on the current general claim of a factor of 2 of gain through software processing, a potential C-SPECT-II system can cut acquisition time at least by a combined factor of 8 and support sub-2-minute acquisition without compromising current image quality. With the inherently low noise data acquired, the frame rate can likely be pushed further. For a slightly reduced image quality, even the C-SPECT-I could provide enough SGE for sub-2-minute dynamic cardiac SPECT imaging.

Regarding the potential cost of a C-SPECT-I system, we believe it has the potential to be comparable to and competitive with current general-purpose dual-head clinical SPECT systems. This is because the cost of a SPECT system is mainly determined by the two major hardware components – the detector system and the mechanical system. It is the reduced cost of the mechanical system that makes C-SPECT-I attractive and could offset the higher cost of a new detector system.

V. CONCLUSION

We have presented the design concepts of the C-SPECT platform to achieve high performance imaging based on an optimized system geometry for efficient and effective detection of photons. We have introduced two new concepts – PIV and collimator exchange – and their ways of implementation that allow adaptive imaging to be realized clinically. We have also presented versatile and flexible sampling schemes for different modes of imaging and described ways to implement these concepts in a basic C-SPECT-I system. Based on a tentative model of C-SPECT-I, we have illustrated how it achieves high SGE and presented its basic sampling schemes. The proposed C-SPECT-I system can double the SGE of cardiac imaging from that of conventional dual-head SPECT systems. This increased SGE can be used to suit various imaging situations and achieve improved performance. We believe, for dynamic imaging, the best performance potential goes to a completely, or near completely, stationary system with the highest SGE and effective sampling. The cost of a C-SPECT-I system could be comparable to and competitive with that of conventional general-purpose dual-head SPECT systems. The most important aspects of the C-SPECT platform, which are hard to put a price tag on, are the improved performance and the potential of expanding cardiac SPECT imaging to new applications and clinical protocols to improve patient care.

Acknowledgments

We thank Dr. J. Halama, Dr. M. Madsen and Dr. J. Oldendick for helpful discussions and editing in preparation of the manuscript. This work is supported in part by USHHS under grant R01HL081414.

References

1. Patton JA, Slomka PJ, Germano G, Berman DS. Recent technologic advances in nuclear cardiology. *J Nucl Cardiol.* 2007; 14(4):501–513. [PubMed: 17679058]
2. Madsen MT. Recent advances in SPECT imaging. *J Nucl Med.* 2007; 48(4):661–673. [PubMed: 17401106]
3. Cardiolite Insert, “Kit for the preparation of Tc-99m sestamibi for injection,” N. Billerica, MA, 2003.
4. Hendel RC, McSherry B, Karimeddini M, Leppo JA. Diagnostic value of a new myocardial perfusion agent, teboroxime (so 30,217), utilizing a rapid planar imaging protocol: preliminary results. *J Am Coll Cardiol.* 1990; 16:855–861. [PubMed: 2229806]
5. Links JM, Frank TL, Becker LC. Effect of differential tracer washout during SPECT acquisition. *J Nucl Med.* 1991; 32(12):2253–2257. [PubMed: 1824556]
6. Dahlberg ST, Weinstein H, Hendel RC, McSherry B, Leppo JA. Planar myocardial perfusion imaging with technetium-99m-teboroxime: Comparison by vascular territory with thallium-201 and coronary angiography. *J Nucl Med.* 1992; 33(10):1783–1788. [PubMed: 1403145]
7. Chiao PC, Ficaro EP, Dayanikli F, Rogers WL, Schwaiger M. Compartmental analysis of technetium-99m-teboroxime kinetics employing fast dynamic SPECT at rest and stress. *J Nucl Med.* 1994; 35(8):1265–1273. [PubMed: 8046477]
8. Smith M, Gullberg GT, CPE. Experimental verification of technetium 99m-labeled teboroxime kinetic parameters in the myocardium with dynamic single-photon emission computed tomography: reproducibility, correlation to flow, and susceptibility to extravascular contamination. *J Nucl Cardiol.* 1996; 3(2):130–42. [PubMed: 8799238]

9. Berman D, Ziffer J, Gambhir S, et al. D-SPECT: A novel camera for high speed quantitative molecular imaging: Initial clinical results. *J Nucl Med.* 2006; 47:131P.
10. Segars, WP.; Mok, SP.; Tsui, BMW. Investigation of equal magnitude respiratory gating in quantitative myocardial SPECT; *Proc IEEE Nuclear Science Symposium Conference Record*; 2006. p. 2107-2110.
11. Tsui MW, Gullberg GT, Edgerton ER, Ballard JG, Perry JR, McCartney WH, Berg J. Correction of nonuniform attenuation in cardiac SPECT imaging. *J Nucl Med.* 1989; 30(4):497–507. [PubMed: 2786944]
12. Ficaro EP, Fessler JA, Shreve PD, Kritzman JN, Rose PA, Corbett JR. Simultaneous transmission/emission myocardial perfusion tomography: Diagnostic accuracy of attenuation-corrected ^{99m}Tc -sestamibisingle-photon emission computed tomography. *Circulation.* 1996; 93:463–473. [PubMed: 8565163]
13. Bailey DL. Transmission scanning in emission tomography. *Eur J Nucl Med.* 1998; 25(7):774–87. [PubMed: 9662601]
14. Tang HR, et al. Implementation of a combined x-ray ct-scintillation camera imaging system for localizing and measuring radionuclide uptake: experiments in phantoms and patients. *IEEE Trans Nucl Sci.* 1999; 46(3):551–557.
15. Corbett JR, Ficaro EP. Clinical review of attenuation-corrected cardiac SPECT. *J Nucl Cardiol.* 1999; 6(1):54–68. [PubMed: 10070841]
16. Hendel RC, et al. Multicenter clinical trial to evaluate the efficacy of correction for photon attenuation and scatter in SPECT myocardial perfusion imaging. *J Nucl Med.* 1999; 99:2742–2749.
17. Zaidi H, Hasegawa B. Determination of the attenuation map in emission tomography. *J Nucl Med.* 2003; 44(3):291–315. [PubMed: 12571222]
18. Hendel RC, Corbett JR, Cullom SJ, DePuey EG, Garcia EV, Bateman TM. The value and practice of attenuation correction for myocardial perfusion SPECT imaging: A joint position statement from the american society of nuclear cardiology and the society of nuclear medicine. *J Nucl Cardiol.* 2002; 9(1):135–43. [PubMed: 11845140]
19. Heller GV, et al. American society of nuclear cardiology and society of nuclear medicine joint position statement: Attenuation correction of myocardial perfusion SPECT scintigraphy. *J Nucl Cardiol.* 2004; 11(1):229–230. [PubMed: 15052255]
20. Chang W, Liang H, Liu J. Design concepts and potential performance of MarC-SPECT - a high-performance cardiac SPECT system. *J Nucl Med.* 2006; 47:190P.
21. Chang W, Ordonez C, Liang H, Holcomb M, Li Y, Liu J. C-SPECT/CT-2: Design concepts and performance potential. *J Nucl Med.* 2008; 49:124P.
22. Rogers WL, et al. SPRINT II: A second generation single photon ring tomograph. *IEEE Trans Med Imaging.* 1988; 7(4):291–297. [PubMed: 18230481]
23. Wieczorek H, Goedicke A. Analytical model for pixellated SPECT detector concepts. *IEEE Trans Nucl Sci.* 2006; 53(3):1102–1112.
24. Metzler SD, Accorsi R, Novak JR, Ayan AS, Jaszczak RJ. On-axis sensitivity and resolution of a slit-slat collimator. *J Nucl Med.* 2006; 47(11):1884–1890. [PubMed: 17079823]
25. Barrett HH, Eskin JD, Barber HB. Charge transport in array of semiconductor gamma-ray detectors. *Phys Rev Lett.* 1995; 75(1):156–159. [PubMed: 10059139]
26. He Z, Knoll GF, Wehe DK, Miyamoto. Position sensitive single carrier CdZnTe detectors. *Nucl Instrument Methods.* 1997; A388:180–185.
27. Bennett PR, Shah KS, Klugerman M, Squillante MR. High efficiency pixellated CdTe detector. *Nucl Instrum Methods A.* 1997; 392(1):260–263.
28. James RB, et al. Material properties of large-volume cadmium zinc telluride crystals and their relationship to nuclear detector performance. *J Elec Mat.* 1998; 27(6):788–799.
29. Pani R, et al. Recent advances and future perspectives of gamma imagers for scintimammography. *Nucl Instrum Methods A.* 2006; 569(2):296–300.
30. Wojcik R, Majewski S, Kross B, Steinbach D, Weisenberger AG. High spatial resolution gamma imaging detector based on a 5" diameter R3292 Hamamatsu PSPMT. *IEEE Trans Nucl Sci.* 1998; 45(3):913–928.

31. Patt BE, Iwanczyk JS, Tull CR, Wang N, Tornai MP, Hoffman EJ. High resolution CsI(Tl)/Si-PIN detector development for breast imaging. *IEEE Trans Nucl Sci.* 1998; 45(3):2126–2131.
32. Wong WH, Li H, Uribe J, et al. Feasibility of a High-Speed Gamma-Camera Design Using the High-Yield-Pileup-Event-Recovery Method. *J Nucl Med.* 2001; 42:624–632. [PubMed: 11337552]
33. Chang W, Liu J, Yu D, Loncaric S. A cylindrical geometry for cardiac SPECT imaging. *IEEE Trans Nucl Sci.* 1996; 43(4):2219–2224.
34. Hudson HM, Larkin RS. Accelerated image reconstruction using ordered subsets of projection data. *IEEE Trans Med Imag.* 1994; 13(4):601–609.
35. Manglos SH. Truncation artifact suppression in cone-beam radionuclide transmission ct using maximum likelihood techniques: evaluation with human subjects. *Phys Med Biol.* 1992; 37(2): 549–562. [PubMed: 1565690]
36. Kadrmas DJ, Jaszczak RJ, McCormick JW, Coleman RE, Lim CB. Truncation artifact reduction in transmission CT for improved SPECT attenuation compensation. *Phys Med Biol.* 1995; 40(6): 1085–1104. [PubMed: 7659732]
37. Jaszczak RJ, Gilland DR, McCormick JW, Scarfone C, Coleman RE. The effect of truncation reduction in fan beam transmission for attenuation correction of cardiac SPECT. *IEEE Trans Nucl Sci.* 1996; 43(4):2255–2262.
38. Gregoriou GK, Tsui BMW, Gullberg GT. Effect of truncated projections on defect detection in attenuation-compensated fanbeam cardiac SPECT. *J Nucl Med.* 1998; 39(1):166–175. [PubMed: 9443757]
39. Yamaguchi C, Yamamoto T, Terada H, Akisada M. Effect of tungsten absorption edge filter on diagnostic x-ray spectra, image quality and absorbed dose to the patient. *Phys Med Biol.* 1983; 28(3):223–232. [PubMed: 6844401]
40. Conwell R, Kindem J, Bai C, Maddahi J. A novel cardiac spect system with x-ray-based attenuation correction using the same solid state detectors for both emission and transmission scans. *J Nucl Med.* 2008; 49:64P.
41. Anger, HO. Instrumentation in Nuclear Medicine. New York: Academic; 1967. p. 485-552.ch. Radioisotope cameras
42. Sorensen, JA.; Phelps, ME. Physics in Nuclear Medicine. 2. Orlando, FL: Grune & Stratton; 1987.
43. Novak JR, Ayan AS, Accorsi R, Metzler SD. Verification of the sensitivity and resolution dependence on the incidence angle for slit-slat collimation. *Phys Med Biol.* 2008; 53(4):953–966. [PubMed: 18263951]
44. Chang W, Liang H, Liu J. Assessing Geometric efficiency of cardiac SPECT systems. *J Nucl Med.* 2006; 47:190P.
45. Joseph PM, Shultz RA. View sampling requirements in Fan-beam computed tomography. *Med Phys.* 1980; 7:692–702. [PubMed: 7464713]
46. Devi, S.; Clackdoyle, R.; Christian, PE. Torso and heart dimensions of cardiac SPECT patients. Proc. IEEE Nuclear Science Symposium Conference Record; New Mexico. 1997. p. 1087-1091.
47. Wackers FJ. Artifacts in planar and SPECT myocardial perfusion imaging. *Am J Card Imaging.* 1992; 6(1):42–57. 2004. [PubMed: 10147614]
48. Abufadel A, Eisner RL, Schafer RW. Differences due to collimator blurring in cardiac images with use of circular and elliptic camera orbits. *J Nucl Cardiol.* 2001; 8(4):458–65. [PubMed: 11481568]
49. O'Connor MK, Hruska CB. Effect of tomographic orbit and type of rotation on apparent myocardial activity. *Nucl Med Comm.* 2005; 26(1):25–30.
50. Ahmed SU, et al. Comparison of gated sestamibi myocardial perfusion imaging between upright solid state and standard supine gamma systems [abstract]. *J Nucl Cardiol.* 2004; 11(4):S14–S15.
51. Lewin H, DePuey E, Hyun M. Upright vs. supine imaging – a clinical comparison. *J Nucl Med.* 2004; 11(4):S6.
52. Cherla A, et al. Technical and clinical evaluation of upright SPECT myocardial perfusion imaging. *J Nucl Med.* 2005; 46:9P.
53. Rogulski MM, Barber HB, Barrett HH, Shoemaker RL, Woolfenden JM. Ultra-high-resolution brain SPECT imaging: simulation results. *IEEE Trans Nucl Sci.* 1993; 40(4):1123–1129.

54. Funk T, Kirch DL, Koss JE, Botvinick E, Hasegawa BH. A novel approach to multipinhole SPECT for myocardial perfusion imaging. *J Nucl Med*. 2006; 47:595–602. [PubMed: 16595492]
55. Steele PP, Kirch DL, Koss J. Comparison of Simultaneous Dual-Isotope Multipinhole SPECT with Rotational SPECT in a Group of Patients with Coronary Artery Disease. *J Nucl Med*. 2008; 49:1080–1089. [PubMed: 18552149]
56. Barrett HH, et al. Adaptive SPECT. *IEEE Trans Med Imag*. 2008; 27(6):775–788.
57. Yu DC, Chang W, Pan TS, Loncaric S. A study of reconstruction accuracy for a cardiac SPECT system with multi-segmental collimation. *IEEE Trans Nucl Sci*. 1997; 44:1403–1408.
58. Jaszczak RJ, Floyd CE, Manglos SH, Greer KL, Coleman RE. Cone beam collimation for single photon emission computed tomography: Analysis, simulation, and image reconstruction using filtered backprojection. *Med Phys*. 1986; 13(4):484–489. [PubMed: 3488496]
59. Hawman, EG.; Hsieh, J. Astigmatic collimator. US Patent. 4,670,657. Jun 2. 1987
60. Jaszczak RJ, Greer KL, Coleman RE. SPECT using a specially designed cone beam collimator. *J Nucl Med*. 1988; 29(8):1398–405. [PubMed: 3261333]
61. Gullberg GT, Christian PE, Zeng GL, Datz FL, Morgan HT. Cone beam tomography of the heart using single-photon emission-computed tomography. *Invest Radiol*. 1991; 26(7):681–8. [PubMed: 1885277]
62. Gullberg GT, Zeng GL, Datz FL, Christian PE, Tung CH, Morgan HT. Review of convergent beam tomography in single photon emission computed tomography. *Phys Med Biol*. 1992; 37(3): 507–534. [PubMed: 1565688]
63. Li J, et al. An evaluation of lesion detectability with cone-beam, fanbeam and parallel-beam collimation in SPECT by continuous ROC study. *J Nucl Med*. 1994; 35(1):135–140. [PubMed: 8271034]
64. Gullberg GT, Zeng GL. Cardiac single-photon emission-computed tomography using combined cone-beam/fan-beam collimation. *IEEE Trans Nucl Sci*. 2005; 52(1):143–153.
65. Frey EC, Tsui BMW. Modeling the scatter response function in inhomogeneous scattering media for SPECT. *IEEE Trans Nucl Sci*. 1994; 41(4):1585–1593.
66. Haynor DR, Kaplan MS, Miyaoka RS, Lewellen TK. Multi-window scatter correction techniques in single-photon imaging. *Med Phys*. 1995; 22(12):2015–2024. [PubMed: 8746706]
67. Digirad dedicated cardiac SPECT. [Online]. Available at website: <http://www.digirad.com>
68. Juni, JE. Single photon emission computed tomography system. US Patent. 7,105,825. Sep. 2006
69. Spectrum dynamics. D-SPECT cardiac imaging system. [Online]. Available at website: <http://www.spectrum-dynamics.com/products/d-spect.html>
70. Lohmann AW, Dorsch RG, Mendlovic D, Zalevsky Z, Ferreira C. Space-bandwidth product of optical signals and systems. *J Opt Soc Am A*. 1996; 13(3):470–473.
71. Barrett, HH.; Hunter, WCJ. Small-Animal SPECT Imaging. Springer; 2005. p. 9-48.ch. Detectors for small-animal SPECT I

Fig. 1. Conceptual rendering of a C-SPECT-I system. A C-shaped stationary gantry surrounds the left front thorax of a patient sitting upright on a Smart-chair. A transmission source, parked in the back of the gantry can move out for Tct imaging.

Fig. 2.

A transverse cross-section of C-SPECT-I, viewed downward from the top, shows a C-shaped detector-arc and a similar shaped slit-slat collimator system. The latter conforms to the gantry's C-arc. Between the slit-plate and the detector-arc is the slat-stack. A large number of transverse non-overlapped projections of the heart can be imaged simultaneously through the slits.

Fig. 3. Cross-section diagrams of a single slit of a slit-slat collimator in front of its detector: (a) transverse view, (b) longitudinal view. The slit-plate and slat-stack are not physically connected. Note the slit-guides on both sides of the slit.

Fig. 4.

Two predetermined imaging volumes (PIVs) are shown with different sizes and locations. One is 15-cm (heavy dash) and the other is 18-cm (light dash) in diameter, in a large (40x 30 cm) and extra-large ellipses representing large patient cross-sections.

Fig. 5.

Scout imaging, on C-SPECT-I, can be performed with a special slit-plate (with 5–7 slits) designed for a large PIV (about 32–44 cm in diameter) that contains the heart. A slit-plate exchange mechanism is embedded in the gantry. Multiple semi-flexible slit-plates are connected and looped around the detector system by a conveyor belt (dashed line). A stepper-motor turns the conveyor-belt and brings a selected slit-plate up to the front for imaging.

Fig. 6.

Tct imaging, on C-SPECT-I, can be conducted with patient rotating when a special collimator (V-blinds) is fixed in place. A transmission source is positioned behind the patient. A 50 cm diameter PIV for Tct is shown as the dotted circle.

Fig. 7. A conventional Dual-head SPECT model for patient imaging. The 2 detectors with collimators are indicated by the two solid rectangles at their initial positions and then by dashed rectangles that progressively change position as detectors rotate.

Fig. 8.

a) The digital heart-body phantom, b) the center quarter of the previous imaging matrix. Reconstructed images with c) 1 set (no rotation), d) 2 sets (1, 10° rotation), e) 3 sets (2, 7° rotations), and f) 4 sets (3, 5° rotations) of projections.

Fig. 9.
Reconstructed images with the three ranges of angular sampling: a) 180°, b) 210° and c) 240°

Fig. 10. Horizontal profiles through the center of the three reconstructed images shown in Fig. 9. These 3 profiles, with offset vertical scale, show little difference within the 15 cm PIV, which is located between pixel numbers 40 to 90.

Fig. 11. Sum-of-squared-error, inside the 15 cm PIV, is plotted as a function of angular-sampling range

Fig. 12.

Sampling pattern in Sinogram space for a 12-slit C-SPECT-I with 3 sets of projections (with two 7° rotations). The abscissa is the radial positions of the projected data. The ordinate is the corresponding projection angles.

TABLE I

Imaging Requirements and Basic Parameters for a 12-slit C-SPECT-I model

Symbol	Parameter	Value (mm)
R_{st}	Transverse sampling resolution at center of PIV	11.0
R_{it}	Transverse intrinsic resolution of detector	3.0
d	Slit width	3.0
R_{sl}	Longitudinal sampling resolution at center of PIV	11.0
R_{il}	Longitudinal intrinsic resolution of detector	3.7
t	Slit thickness	0.2
h	Distance between adjacent slats	3.0

TABLE II

Calculation of System Geometric Efficiency for a 12-slit C-SPECT-I model

slit no.	1	3	5	7	9	11
distance from the slit to the center of PIV, b (mm)	254	199	161	158	186	223
radial distance from the slit to the detector-arc, L_r (mm)	137	107	87	86	101	121
effective radial length of the slit, L_e (mm)	114	89	72	71	84	100
GE per slit ($\times 10^{-5}$)	2.57	4.21	6.46	6.71	4.83	3.33
Total SGE = Sum of all GE ($\times 10^{-5}$)	56.4					

TABLE III

Calculation of System Geometric Efficiency for a Dual-Head SPECT model

Rotational step No.	1	9	17	25	33	41	49	57	Ave.
distance from collimator to the center of PIV (<i>mm</i>)	201	162	138	130	141	165	188	197	
collimator resolution (hole length = 23.1 mm)(<i>mm</i>)	12.23	10.24	8.94	8.51	9.09	10.35	11.55	12.05	10.35
sampling resolution in both <i>t</i> and <i>l</i> directions (<i>mm</i>)	12.51	10.88	9.67	9.28	9.81	10.99	12.13	12.61	11.0
GE per head ($\times 10^{-5}$)									14.1
Total SGE = GE $\times 2$ ($\times 10^{-5}$)									28.2

TABLE IV

Comparison of SGEs between the two system models

no. of slit/ISR	Dual-head model	C-SPECT-I model		
		11-slit/3.3 mm	12-slit/3.0 mm	13-slit/2.7 mm
SGE ($\times 10^{-5}$)	28.2	53.9	56.4	58.0
Ratio (SGEI)	1.00	1.91	2.00	2.05