Size-dependent scanning parameters (kVp and mAs) for photon-counting spectral CT system in pediatric imaging: simulation study

Han Chen^{a,*}, Mats Danielsson^a, Cheng Xu^a

^a*Dept. of Physics, Royal Institute of Technology (KTH), AlbaNova University Center, 106 91 Stockholm, Sweden*

Abstract

We are developing a photon-counting spectral CT detector with small pixel size of $0.4 \times 0.5 \text{ mm}^2$, offering a potential advantage for better visualization of small structures in pediatric patients. The purpose of this study is to determine the patient size dependent scanning parameters (kVp and mAs) for pediatric CT in two imaging cases: adipose imaging and iodinated blood imaging.

Cylindrical soft-tissue phantoms of diameters between 10-25 cm were used to mimic patients of different ages from 0-15 y. For adipose imaging, a 5-mm-diameter adipose sphere was assumed as an imaging target, while an iodinated blood sphere of 1 mm in diameter was assumed in the case of iodinated imaging. By applying the geometry of a commercial CT scanner (GE LightSpeed VCT), simulations were carried out to calculate the detectability index, d'^2 , with tube potentials varying from 40 to 140 kVp. The optimal kVp for each phantom in each imaging case was determined such that the dose-normalized detectability index, d'^2/dose , is maximized. With the assumption that image quality in pediatric imaging is required the same as in typical adult imaging, the value of mAs at optimal kVp for each phantom was selected to achieve a reference detectability index that was obtained by scanning an adult phantom (30 cm in diameter) in a typical adult CT procedure (120 kVp and 200 mAs) using a modeled energy-integrating system.

For adipose imaging, the optimal kVps are 50, 60, 80, and 120 kVp, respectively, for phantoms of 10, 15, 20, and 25-cm in diameter. The corresponding mAs values required to achieve the reference detectability index are only 9%, 23%, 24%, and 54% of the mAs that is used for adult patients at 120 kVp, for 10, 15, 20, and 25-cm-diameter phantoms, respectively. In the case of iodinated imaging, a tube potential of 60 kVp was found optimal for all phantoms investigated, and the mAs values required to achieve the reference detectability index are 2%, 9%, 37%, and 109% of the adult mAs. The results also indicate that with the use of respective optimal kVps, the photon-counting spectral system offers up to 30% higher d'^2/dose than the modeled energy-integrating system for adipose imaging, and 70% for iodinated imaging.

Keywords: photon counting, spectral CT, silicon-strip detector, pediatric imaging, detectability index

1. Introduction

CT imaging has the advantage of high speed which is important in acute settings (e.g. emergency rooms). Also with the high speed children do not need to be sedated to avoid motion artifacts, as for magnetic resonance imaging (MRI), reducing risk and cost associated to this process. In 2006, 4 million children were examined with CT in the USA [1] and today well over 10 million children is probably examined worldwide[2]. Since children have more rapidly dividing cells than adults and have longer life expectancy[3, 4], the odds that they will develop cancers from x-ray radiation may be higher than for adults. A study has estimated that the lifetime radiation risk for children is 2-3 times greater than that to adults[5]. Moreover, a reduced cognitive function in adulthood as a result of low radiation dose to the brain in infancy has also been reported[6]. Hence, it is of particular importance to keep

the radiation in pediatric CT as low as reasonably achievable (ALARA).

Initial effort to reduce radiation risk for pediatric CT is to modulate the value of milliamperes-seconds (mAs) based on patient weight or size[7, 8, 9, 10, 11]. While some hospitals experientially chose tube current settings based on patient weight[10], Boone et al[11] physically determined size-dependent mAs by performing a phantom study. The results showed that with the same CT operation at 120 kVp, pediatric patients of 15 cm in diameter only required 5.4% of the tube current used for adults while image quality was maintained. Aside from the modulation of mAs, there is a growing interest recently in optimizing tube potentials for pediatric CT. The general conclusion is that the use of low tube potentials can reduce dose with preserved image quality[12, 13, 14, 15, 16]. A phantom study by Siegel et al[14] shows that decreasing tube voltage from 120 to 80 kVp can reduce dose in pediatric contrast-enhanced CT without compromising image quality. The dose reduction was also found in low-contrast imaging by

*Corresponding author

Email address: hanchen@kth.se (Han Chen)

Figure 1: (a) Photograph of a single detector module. x-rays are incident from the top of the detector module and the ASICs are located on the right side protected by a thick tungsten slab. (b) Illustration of a full CT detector with a two-layer design.

reducing the tube voltage from 120 to 90 kVp, as reported by Funama et al[15]. The underlying principle of the improvement at low tube potentials is that low-energy x-rays contains more contrast information than high-energy x-rays, particularly in contrast-enhanced imaging where the attenuation coefficients of contrast agents are significantly increased at low energies. Although lowering tube potential is an effective way of reducing radiation dose in pediatric CT, pediatric patients were scanned by many hospitals with the same scanning protocols as adults[17]. The main challenge is that while radiologists have rich experience in the tube current settings for conventional 120 kVp operation, they are uncertain about which reference tube current should be implemented at a reduced tube potential[18]. In addition, because low-energy x-rays are more likely to be attenuated, the tube currents required for children with large sizes perhaps exceed the maximum current achievable in x-ray tubes.

We are developing a photon-counting spectral (PCS) detector for CT applications[19, 20, 21, 22, 23]. Compared to energy-integrating (EI) detectors that are commonly used in commercial CT systems, PCS detectors are capable of measuring energies of individual x-rays and sorting them into different energy bins. Spectral information obtained by the PCS detectors can be used to improve signal-difference-to-noise ratio (SDNR) by applying energy-weighting schemes[24, 25], a potential trade-off for lower radiation dose. Moreover, the pixel size ($0.5 \times 0.4 \text{ mm}^2$) of our PCS detector is only around one-fifth as small as those of typical CT detectors, thus allowing for better visualization of small structures in pediatric patients. The purpose of this study is to determine patient size dependent scanning parameters, kVp and mAs, for pediatric CT based on our PCS detector. Two kinds of imaging cases are investigated in the study: adipose imaging and iodinated blood imaging. While the former imaging case is representative of low-contrast tasks in abdomen examinations, the latter case is a high-contrast task commonly encountered in clinics. The image quality is quantified in terms of detectability index (d'^2), a figure of merit that takes into account the spatial frequency dependence of signal

and noise.

2. Materials and methods

2.1. Description of photon-counting spectral detector

A photography of a single detector module is shown in Fig.1 (a). The detector module is fabricated on a high-resistivity n-type silicon substrate with a thickness of 0.5 mm, and consists of 50 detector strips each with a pitch of 0.4 mm. As a result, a pixel size of $0.5 \times 0.4 \text{ mm}^2$ is given for each detector strip by orienting the module with its edge directed towards the x-ray beam (i.e., edge-on). The active absorption path along the x-ray incident direction is 30 mm long with the aim of providing a high detection efficiency. In order to overcome the problem of high photon fluxes, the detector strip is subdivided into 16 segments along the x-ray incident direction. The segment length is exponentially increased, providing a nearly uniform count rate over all segments. Each segment is connected to an individual electronic readout channel on an application specific integrated circuits (ASIC) bonded directly to the right side of the silicon substrate. The ASICs serve to amplify and shape the current pulses produced by x-ray interactions with the pulse height proportional to the amount of the deposited energy, and the resulting pulses are compared with eight energy thresholds and then sorted into one of the eight energy bins formed by two neighboring energy thresholds. More detailed information about the ASIC can be found in earlier publications[20, 19].

Fig.1 (b) illustrates the geometry of a full CT detector being assembled, which consists of a large number (1500-2000) of detector modules. Each module is aligned with its front edge pointing towards x-ray source. To facilitate the cooling and mounting of electronics, the detector modules are stacked in two different layers, with the lower detector layer offset by one module thickness along the x-axis relative to the upper layer. The backside of each detector module is coated by a $50\text{-}\mu\text{m}$ -thick sheath of tungsten to reduce internal scatter radiation between different detector modules, yielding a geometrical efficiency of around 0.95. In order to reject the scatter radiation from objects,

Figure 2: Schematic of the simulation setup with a full CT detector. Phantoms are irradiated by a fan beam with a width of 1 cm along the z-axis at the isocenter.

the tungsten sheaths at the upper detector layer are extended by 2.5 cm towards x-ray source, acting as a one-dimensional anti-scatter grid.

A schematic of the simulation setup used in the study is shown in Fig.2. The simulations are performed based on the CT geometry of a commercial CT scanner (GE LightSpeed VCT). The source-to-isocenter distance is 541 mm and the source-to-detector distance is 949 mm, giving a geometric magnification 1.75. The gantry rotation time is assumed to be 0.5 s and the total number of basis projections acquired over one rotation is assumed to be 2400. Four soft-tissue cylinders with diameters of 10, 15, 20, and 25 cm are placed at the isocenter to mimic a new-born baby, a 5-y patient, a 10-y patient, and a 15-y patient, respectively, and an additional phantom of 30 cm in diameter is studied to mimic an adult patient. The axial lengths of all the phantoms are 14 cm. For adipose imaging, a 5-mm-diameter adipose sphere was assumed as an imaging target, while an iodinated blood sphere with 1-mm diameter and 5 mg/ml iodine concentration was assumed in the case of iodinated imaging. The phantoms are irradiated by a x-ray fan beam with a width of 1 cm along the z-axis at the isocenter. The x-ray tube equipped in GE light speed VCT is Performix Pro 100 which has a target angle of 7 degree and a focal spot size of 0.6×0.7 mm²[26]. Aside from an inherent flat filter equivalent to 0.43-cm-thick aluminum (Al), the GE LightSpeed system provides three types of bowtie filters—small, medium and large—which cover the full range of patients sizes, from pediatric/head to adult patients. The geometry of the three bowtie filters were measured using the method described in Appendix A. The resulting equivalent Al thicknesses of the bowtie filters in Fig.3 as a function of fan beam angle are shown. In this study, the small bowtie filter is used for all the pediatric phantoms, and the large bowtie filter for the adult phantom.

2.2. Figure of merit

The detectability index, which describes spatial frequency weighted SDNR based on the linear ideal observer model, is applied for the evaluation of image quality. In the published literatures[27, 28], there are two different methods of determining the detectability index d'^2 for CT imaging systems: the 3D volumetric detectability index

Figure 3: Equivalent-aluminum (Al) thickness of the bowtie filter as a function of fan angle, measured in GE LightSpeed VCT. The fan angle refers to the angle relative to the central x-ray path in the x-y plane.

and the more familiar 2D detectability index in the central slice of the reconstructed image. It is an open question which one of the two methods is better to approximate a human observer[29]. In this work, the latter method is applied. Let $\Delta\mu$ denote the peak difference in attenuation coefficient between two hypotheses: h_1 for the presence of an imaging target and h_0 for the absence of the target. Then, d'^2 of the target in the reconstructed imaging is written as:

$$d'^2 = (\Delta\mu)^2 V^2 \int \int \frac{|\int \text{MTF}_{3D}(f_x, f_y, f_z) W(f_x, f_y, f_z) df_z|^2}{\int \text{NPS}_{3D}(f_x, f_y, f_z) df_z} df_x df_y \quad (1)$$

where V is the volume of the imaging target. The task function W is obtained by the Fourier transform of the difference between two hypotheses, h_0 and h_1 . Assuming that the imaging target has a thickness of d_t , $\Delta\mu$ for a rotationally symmetrical phantom can be determined:

$$\Delta\mu = |\ln \frac{\bar{q}^0}{\bar{q}^t}| \times \frac{1}{d_t}, \quad (2)$$

with $\bar{q}^m$ being the expected signal under hypothesis h_m ($m \in [0, 1]$) for x-ray path through the central part of the phantom.

The terms MTF_{3D} and NPS_{3D} in Eq (1) represent the 3D modulation transfer function and noise power spectrum in the reconstruction domain, respectively, with f_x , f_y , and f_z being the spatial frequencies in the x-, y-, and z-axis. Using the method presented by Tward and Siewerdsen[30], 3D MTF and NPS in the image reconstruction domain can be calculated by taking the 2D MTF(u, v) and NPS(u, v) in the projection domain into the equations below:

$$\begin{aligned} \text{MTF}_{3D}(f_r, f_z) &= \frac{1}{u} \text{MTF}_{2D}(u/M, v/M) T_r T_{apo} T_{interp} \\ \text{NPS}_{3D}(f_r, f_z) &= \frac{\pi M}{(\bar{q}^0)^2 N_p u} \text{NPS}_{2D}(u/M, v/M) T_r^2 T_{apo}^2 T_{interp}^2 \end{aligned} \quad (3)$$

where M is the system geometric magnification, N_p is the total number of basis projections over one CT rotation, u and v represent the spatial frequencies of the projection domain in the x- and z-axis, respectively, and f_r is the radial frequency in the axial plane (x-y plane), given as $\sqrt{f_x^2 + f_y^2}$. The terms T_r , T_{apo} and T_{interp} refer to the filters applied in the filtered back projection algorithm. T_r is the ramp filter, given by $|u|$. T_{apo} is apodization filter to limit high-frequency noise, determined by a Hanning window function. The noise power spectrum through the interpolation process is described by a sinc function, $T_{interp} = \text{sinc}^2(\pi u a_x) \text{sinc}^2(\pi v a_z)$, with a_x and a_z being the pixel widths along the x- and z-axis in the projection domain, respectively.

To determine the most dose efficient tube potential, the detectability index is normalized by the corresponding dose delivered to the phantom, d'^2/dose , a figure of merit independent of radiation exposure. Then, the optimal kVp for each phantom in each imaging case is selected such that d'^2/dose is maximized.

The value of mAs at the optimal kVp is selected as follows. For each imaging case, we define a reference detectability index d'_{ref}^2 that is obtained by scanning the adult phantom in a typical adult CT procedure (120 kVp and 200 mAs) using an energy-integrating (EI) system modeled in section 2.4. With the assumption that image quality for pediatric imaging is required the same as for adult imaging, the corresponding mAs for each pediatric phantom is then determined to achieve a detectability index equal to d'_{ref}^2 . Since the values of mAs in adult CT procedures might vary substantially due to different CT scanners and imaging tasks, the calculated mAs values for the investigated pediatric phantoms are normalized by 200 mAs to cancel out the scanner- and task-specific factors.

2.3. Theoretical Model of photon-counting spectral system

2.3.1. 2D projection MTF

We define point spread function (psf) as the expected signal distribution over the projection domain as a result

of a parallel x-ray beam incident on a pixel. For a photon-counting spectral (PCS) system with N energy bins, a weighted projection image is obtained by weighting the counts registered by each energy bin. Thus, the total psf is a linear combination of psfs of N energy bins.

$$\text{psf} = \sum_{i=1}^N \omega_i \text{psf}_i, \quad (4)$$

where the subscript i denotes the bin number and ω_i refers to the weighting factor assigned to bin i . The resulting presampled detector MTF $_d$ for monochromatic x-rays with energy E is then determined as the zero-frequency normalized Fourier transform of psf. The extension to a broad spectrum is straightforward, given by:

$$\text{MTF}_d(u, v) = \frac{\int \Phi^m(E) D(E) \text{MTF}_d(u, v; E) dE}{\int \Phi^m(E) D(E) dE} \quad (5)$$

where $\Phi^m(E)$ (unit: ph/mm²) refers to the spectrum incident on the detector under hypothesis h_m and $D(E)$ is the energy dependent detection efficiency of the detector, both of which can be obtained by the application of the Beer-Lambert law.

It is well known that the blurring effect caused by focal spot can degrade spacial resolution of an imaging system. By assuming that the intensity of the focal spot in the system follows a rectangular distribution, the final 2D MTF in the detector plane is then determined by:

$$\begin{aligned} \text{MTF}_{2D}(u, v) &= \\ \text{MTF}_d(u, v) &\times |\text{sinc}[(M-1)l_x u] \text{sinc}[(M-1)l_z v]| \end{aligned} \quad (6)$$

with l_x and l_z being the focal spot widths along the x and z direction, respectively.

2.3.2. 2D projection NPS

With the fundamental assumptions that the detector system is linear and shift-invariant and all noise processes are Wide-sense stationary, the projection NPS is determined using the following expression[31]:

$$\text{NPS}_{2D}(u, v) = \frac{a_x a_z}{N_x N_z} < |\mathcal{F}_{x,z}\{\Delta I(x, z)\}|^2 >, \quad (7)$$

where $\Delta I(x, z)$ is the signal difference relative to the expected value in a noise-only image of size $N_x \times N_z$ [i.e., $\Delta I = \bar{I}(x, z) - I(x, z)$]. Let $\Delta I_i(x, z)$ denote the signal difference for bin i image, ΔI can be written in a matrix form:

$$\Delta I(x, z) = \sum_i^N \omega_i \Delta I_i(x, z) = \mathbf{w}^T \Delta \mathbf{I} \quad (8)$$

with $\mathbf{w}$ being the bin weighting vector $[\omega_1, \dots, \omega_N]$. To make the dependence of the NPS on $\mathbf{w}$ clear, Eq 7 is rewritten

ten as:

$$\begin{aligned} \text{NPS}_{2D}(u, v) &= \frac{a_x a_z}{N_x N_z} < |\mathbf{w}^T \mathcal{F}_{x,z} \{\Delta \mathbf{I}(x, z)\}|^2 > \\ &= \mathbf{w}^T \mathbf{NPS} \mathbf{w}, \end{aligned} \quad (9)$$

where $\mathbf{NPS}$ is a $N \times N$ matrix with the elements $\text{NPS}_{ij}(u, v)$ obtained by:

$$\begin{aligned} \text{NPS}_{ij}(u, v) &= \\ &= \frac{a_x a_z}{N_x N_z} < \mathcal{F}_{x,z} \{\Delta I_i(x, z)\}^* \mathcal{F}_{x,z} \{\Delta I_j(x, z)\}^T >. \end{aligned} \quad (10)$$

2.3.3. Optimal bin weighting

For CT imaging, the observed values in different energy bins can be weighted either before log-normalization (projection-based weighting), or after log normalization (image-based weighting). It has been shown that with realistic noise assumptions projection and image-based weighting results in similar SDNR, while the latter method can reduce beam-hardening artifacts[25]. In this work, the bin weights are optimized using the projection-based weighting scheme.

Let $\mathbf{g}$ be a $1 \times N$ column vector $\mathbf{g} = (g_1, g_2, \dots, g_N)^T$, with entries being the outcomes of N energy bins. Under the hypothesis h_m ($m \in [0, 1]$), the expectation value of $\mathbf{g}$ is expressed as $\bar{\mathbf{g}}^m = \langle \mathbf{g} | h_m \rangle$. The squared SDNR between the imaging target and the background is then given by:

$$\text{SDNR}^2 = \frac{(\mathbf{w} \Delta \bar{\mathbf{g}})^2}{\mathbf{w} (K^1 + K^0) \mathbf{w}^T} \quad (11)$$

where $\Delta \bar{\mathbf{g}} = |\bar{\mathbf{g}}^1 - \bar{\mathbf{g}}^0|$ and K^m is the $N \times N$ covariance matrix of $\mathbf{g}$ under hypothesis m , with entries given by:

$$K_{ij}^m = \langle (g_i - \bar{g}_i)(g_j - \bar{g}_j) | h_m \rangle, m = 0, 1. \quad (12)$$

It has been shown that the optimal weighting vector $\mathbf{w}$ that maximizes Eq (11) is determined by[32]:

$$\mathbf{w} = (K^0 + K^1)^{-1} \Delta \bar{\mathbf{g}}. \quad (13)$$

For an ideal case that all incident x-rays are recorded only once by detectors, the outcomes of energy bins are independent of each other and K^m then becomes a diagonal matrix with the diagonal entry K_{ii}^m equal to the expected value of bin i , i.e., $\bar{g}_i^m$. However, this is not the case in practical photon-counting detectors where single x-ray photons might be double counted due to Compton scatter, K-fluorescence effect and charge sharing. The resulting noise events would increase the bin variance K_{ii} if a single x-ray is counted twice in bin i , or introduce non-zero correlation noise K_{ij} if a single x-ray is recorded by two different bins. Hence, these events have to be taken into account in the determination of optimal bin weighting.

Let $\alpha_i(E)$ be the probability of a primary interacting x-ray with original energy E being single registered by bin i ,

and $\beta_{ij}(E)$ the probability that a primary x-ray produces one count in bin i in its original pixel and one more count in bin j in another pixel. The entries of K^m has been determined in Appendix B:

$$K_{ij}^m = \begin{cases} \int a_x a_z \Phi^m(E) D(E) [\alpha_i + \sum_{k=1}^N [\beta_{ik} + \beta_{ki}] + 2\beta_{ii}] dE & i = j \\ \int a_x a_z \Phi^m(E) D(E) [\beta_{ij} + \beta_{ji}] dE & i \neq j \end{cases} \quad (14)$$

The expected value $\bar{g}_i^m$ in bin i is calculated by summing together the primary counts P_i and double counts S_i that are caused by Compton scatter and charge sharing:

$$\begin{aligned} \bar{g}_i^m &= P_i + S_i \\ &= \int a_x a_z \Phi^m(E) D(E) [\alpha_i(E) + \sum_{k=1}^N \beta_{ik}(E)] dE \\ &\quad + \int a_x a_z \Phi^m(E) D(E) \sum_{k=1}^N \beta_{ki}(E) dE. \end{aligned} \quad (15)$$

The optimal weighting vector $\mathbf{w}$ is then determined by substituting Eq (14) and Eq (15) into Eq (13) and the expected value $\bar{q}^m$ in Eq (2) is given by $\mathbf{w}^T \mathbf{g}^m$.

2.4. Theoretical model of energy-integrating system

The energy-integrating (EI) detector is assumed to have unity quantum efficiency, no scatter radiation between pixels and no electronic noise. In contrast to the pixel size of $0.5 \times 0.4 \text{ mm}^2$ for the energy-resolving system, the pixel size of the energy-integrating system is $1.2 \times 1.2 \text{ mm}^2$ with an effective absorption area of $1 \times 1 \text{ mm}^2$. The resulting geometrical efficiency, η_g , is around 0.7 which is a typical value for commercial CT systems. The presampled detector MTF is then determined as the aperture MTF multiplied by the MTF of focal spot :

$$\begin{aligned} \text{MTF}_{2D}(u, v) &= \\ &= |\text{sinc}(a_x u) \text{sinc}(a_z v)| \times |\text{sinc}[(M-1)l_x u] \text{sinc}[(M-1)l_z v]| \end{aligned} \quad (16)$$

with a_x and a_z being 1 mm. The expected value $\bar{q}$ and the projection NPS for the EI system were derived by Siewerdsen et al[33] by applying cascaded model:

$$\bar{q} = \eta_g a_x a_z \bar{\Phi} \lambda_1 \lambda_2, \quad (17)$$

$$\text{NPS}_{2D}(u, v) = \frac{a_x^2 a_z^2}{\eta_g} \left[\frac{\eta \lambda_1^2 \lambda_2^2 \bar{\Phi}}{A_s} + \eta \lambda_1 \lambda_2 \bar{\Phi} \right], \quad (18)$$

where $\bar{\Phi}$ is the average number of x-rays incident on the detector per mm^2 , and λ_1 is the average light quanta per x-ray interaction with the detector, obtained by dividing the mean energy $\bar{E}$ of incident spectrum by the energy W that is required to release a single light photon in detector

Figure 4: Bin psfs for 80-keV x-rays with 8 energy thresholds evenly distributed between 5 and 90 keV.

materials. The detector material in this study is assumed to be $\text{Gd}_2\text{O}_2\text{S:Tb}$ which is commonly used in today's CT and gives a value of 17 eV for W . A_s is the Swank factor accounting for fluctuation of λ_1 due to K-fluorescence escape in the detector material. For simplicity, A_s is assumed to be unity in the study. λ_2 is the factor describing the probability of light quanta being lost during transportation, including the coupling efficiency of light escaping from detector materials and the optical quantum efficiency of photodiodes. Since the values of λ_2 for commercial CT scanners are not available in present literatures, we determined λ_2 to be 0.5, a value derived from the data measured by Luhta et al[34] on a CT prototype,

By applying the same CT geometry as used for the PCS system, the reference detectability index d'_{ref} as defined in section 2.2 is calculated using an adult soft-tissue phantom (30 cm in diameter) and a typical adult procedure (120 kVp and 200 mAs). In the study, we also calculate the detectability indexes resulting from the modeled EI system for the investigated pediatric phantoms and compare to those obtained from the PCS system.

2.5. Simulation

2.5.1. 2D MTF

In order to determine psf_i for the PCS system, Monte Carlo simulations using GATE (Geant4 Application for Tomographic Emission) were performed based on the full CT detector geometry as described in section 2.1. Monochromatic x-rays with energies ranging from 15 to 140 keV were applied in the simulation in steps of 1 keV. For each energy point, one million x-rays uniformly incident on a central pixel were tracked until the energy was totally absorbed or escaped from the whole detector volume. For each x-ray interaction, the interacting position and the corresponding deposited energy were recorded. Since the current version of ASIC does not implement anti-coincidence logic for charge-sharing correction, the effect of charge sharing was

taken into account in the simulation by applying a charge-sharing model developed by Bornefalk et al [35]. In this model, the charge cloud produced by an x-ray is assumed to be spherical Gaussian distributed with a standard deviation, depending on the deposited energy and interaction depth in the detector. The charge leaking to neighboring pixels is simply determined by integrating the charge cloud over the neighboring pixel volumes, disregarding the distortion of electric field between the pixels. Due to the effect of electronic noise, the actual energies detected by a detector pixel might deviate from the deposited energies. This deviation was modeled as a Gaussian distribution with a standard deviation of 1.5 keV measured on a single detector module in earlier work[36]. With the determined detected energies, the interacting events were then assigned into one of 8 energy bins depending on the bin boundaries (i.e., energy thresholds). In this study, the lowest threshold was set to be 5 keV to reject false events produced by electronic noise, while the remaining thresholds were evenly distributed between 5 keV and the maximum x-ray energy of a given incident spectrum.

An example of bin psfs for 80 keV x-rays is shown in Fig. 4. Due to the tungsten shielding between the detector modules, the x-ray re-absorption events along the x-axis are much less than those along the z-axis. For a given broad spectrum, the 2D MTF is determined by taking the MTFs of the monochromatic energies into Eq (5). Fig. 5 shows the 2D MTFs for 80 kVp and 120 kVp. It is clear that for the PCS system, MTF for 80 kVp is slightly better than that of 120 kVp since the amount of Compton scatter in the detector reduces as kVp decreases. Also plotted in Fig. 5 is the result of the EI detector modeled in section 2.4. Due to the larger pixel size, MTF of EI system drops faster than that of PCS system.

It is assumed that the energy responses of all detector pixels are uniform. By reusing the data of the bin psf simulation, the probabilities α_i was calculated as the ratio between the number of x-rays that were single registered

Figure 5: 2D MTF of the PCS system at 80 and 120 kVp for iodinated imaging, in comparison with that of the modeled EI system.(a) MTF($u,0$); (b) MTF($0,v$). A 0.6 mm $\times$ 0.7 mm focal spot was assumed for both systems.

Table 1: Number of primary counts, and number of double counts caused by charge sharing and Compton scatter, in 8 energy bins as a result of 10^6 incident x-rays drawn from 120 kVp spectrum. Also presented are the total bin counts g_i and noise-to-primary ratio γ_i of each bin.

Energy bin (keV)	1 5-20	2 20-35	3 35-50	4 50-65	5 65-80	6 80-95	7 95-110	8 110-120
P_i	2.4e+5	5.2e+4	2.8e+4	8.4e+4	6.2e+4	2.8e+4	1.3e+4	0.2e+4
S_i^{cs}	1.2e+4	4.4e+3	1.0e+3	64	0	0	0	0
S_i^{com}	1.5e+4	7.0e+2	2.2e+3	1.0e+4	3.5e+3	6.8e+2	1.4e+2	2
$g_i = P_i + S_i^{cs} + S_i^{com}$	2.7e+5	5.7e+4	3.2e+4	9.5e+4	6.5e+4	2.9e+4	1.3e+4	0.2e+4
$\gamma_i = (S_i^{cs} + S_i^{com})/P_i$	0.11	0.10	0.11	0.12	0.06	0.02	0.01	0

¹ S_i^{cs} and S_i^{com} denote the double counts due to charge sharing and Compton scatter, respectively.

by energy bin i and the total number of the primary interacting x-rays. It is noted that if an x-ray deposited energy less than the minimum threshold 5 keV in its original pixel and produced one more count in another pixel, we considered it as a single-counted event since it only leads to image blurring and dose not contribute to noise. Similarly, the probability β_{ij} of each monochromatic energy was also calculated in the simulations. For a given incident broad spectrum Φ , the values of α_i and β_{ij} were obtained by averaging the monochromatic results over the spectrum. Table 1 shows the number of primary counts and double counts of 8 energy bins for 120 kVp spectrum, as calculated by Eq (15). It is clear that low-energy bins are severely affected by charge sharing and Compton scatter; the noise-to-primary ratios for bin 1-4 are around 11-12%. Overall, 6.5% and 3.4% of the primary interacting x-rays are double counted due to Compton scatter and charge sharing, respectively.

2.5.2. 2D NPS

To determine the **NPS** matrix in Eq (9) for the PCS system, 1000 noise-only projection images **I** for each monochromatic energy were simulated. For each image, one million incident x-rays uniformly distributed over 50 $\times$ 50 pixels at the center of the detector plane were tracked. Because the re-absorption of scattered x-rays will be underestimated in

the pixels close to the edge of the simulated parallel beam, a region of interest covering the central 40 $\times$ 40 pixels was cropped from each simulated image. The entries of the matrix **NPS** were then calculated through Eq (10). The **NPS** for a given incident spectrum was obtained by averaging the monochromatic results over the incident spectrum. Fig.6 (a) shows the 2D NPS of bin 1 and 2 (i.e., entries NPS₁₁ and NPS₂₂ in **NPS** matrix) at 120 kVp. The non-zero NPS₁₂ in Fig.6 (a) indicates that there exists correlation between bin 1 and 2 which is attributed to the x-rays that are double counted by the two bins. By taking the **NPS** matrix into Eq (9), the total NPS for the PCS system can be computed. Fig.6 (b) shows the resulting NPS normalized by the square of the mean signal, $\bar{q}$, in the pure photon-counting mode (i.e., $\mathbf{w} = 1$). The resulting 2D NPS for the modeled EI system is also presented for comparison, .

2.5.3. Phantom dose

The doses to the phantoms were also determined using Monte Carlo simulation by applying a series of monochromatic x-rays from 5 to 140 keV in 5 keV steps. The divergence of the fan beam was modeled to be just broad enough to cover the phantoms. For each keV and each phantom diameter, one million x-rays were tracked until they were completely absorbed by the phantom, or es-

Figure 6: (a) 2D NPS of bin 1 and 2 for the PCS system at 120 kVp and the correlation noise power spectrum NPS_{12} between the two bins; (b) The corresponding normalized NPS using pure photon-counting mode (i.e., $w = 1$). For comparison, the results of EI system are plotted by crosses.

caped from the phantom volume. The interaction position of each interaction event in the phantom volume and its corresponding deposited energy were recorded. Secondary radiation from the bowtie filter was not included in the simulation, since it contributes little to the phantom dose due to the blocking of pre-collimator in front of the bowtie filter. Fig. 7 shows the radiation dose profile along the z-axis for the 15-cm phantom under 80 keV irradiation. A considerable fraction of energy was deposited outside the directly irradiated area due to scatter radiation. The dose to the phantom for each keV and each phantom diameter was determined as the integral of the radiation dose profile along the z-axis divided by the fan-beam width, which represents the multiple scan average dose (MSAD) of the central phantom volume in helical mode with a pitch equal to 1[37]. The dose for broad spectrum was computed by weighting the monoenergetic results. Since the phantoms used in our study are rotationally symmetrical, the dose received by a phantom per rotation was determined by multiplying the dose per basis projection with the total number of basis projections N_p over one rotation.

It is well known that in practice, MSAD for a given phantom can be estimated by measuring the weighted computed tomography dose index ($CTDI_w$, i.e., the sum of one-third $CTDI$ measured at the center of the phantom and two-thirds the $CTDI$ measured at the edge of the phantom)[38]. The $CTDI_w$ of different commercial scanners are specifically measured using two cylindrical PMMA phantoms of the same length of 14 cm, one for head examination with 16 cm diameter and a second for body examination with 32-cm diameter, defined by FDA[39]. In order to validate the Monte Carlo simulations in our study, we estimated the MSAD by applying the same phantoms and then made a comparison with the values of $CTDI_w$ measured on a GE LightSpeed scanner by manufacturers.

Figure 7: Dose profile along the table movement direction (i.e., z-axis as illustrated in Fig.2) as a result of 10^6 incident x-rays with original energy of 80 keV. The tails of the dose profile are caused by scatter radiation.

3. Results

3.1. 3D MTF and NPS

3D MTF and NPS in the reconstruction image domain are computed by taking the simulated 2D MTF and NPS into Eq (3). Fig. 8 (a) shows the radial frequency ($f_r = \sqrt{f_x^2 + f_y^2}$) dependent 3D MTF with $f_z = 0$ at 120 kVp, for PCS system by applying the pure photon-counting (p.c.) weighting and optimal energy weighting (e.w.). Also shown are the results of the modeled EI system for comparison. Fig. 8 (b) shows the corresponding results of 3D NPS for both systems.

3.2. Phantom dose

Fig.9 (a) shows the phantom diameter dependence of dose (unit: Gy/mAs) for 60, 80, and 120 kVp. The dose is gradually decreased as the phantom diameter increases. In order to validate the Monte Carlo simulations, comparisons with the values of $CTDI_w$ [26] measured on a GE

Figure 8: (a) MTF_{3D} as a function of radial frequency $f_r = \sqrt{f_x^2 + f_y^2}$ with $f_z = 0$, for the PCS system. The incident spectrum of 120 kVp is assumed filtered by a 15-cm phantom. The results from two different weighting modes, pure photon-counting (p.c.) mode and optimal energy-weighting mode, are presented, together with the results of the modeled EI system; (b) The corresponding results of 3D NPS for both systems.

Figure 9: (a) Dose (mGy/mAs) as a function of soft-tissue phantom diameter for 60, 80, and 120 kVp. (b) Dose resulting from two PMMA phantoms as specified by FDA, one for head examination and the other for body examination, as a function of kVp, in comparison with the values of CTDI_w measured on a GE LightSpeed VCT by manufacturers[26].

Figure 10: Dose-normalized detectability index, d^2/dose , as a function of kVp for adipose imaging. The results of the PCS system are plotted as solid and dashed lines for pure photon-counting (p.c.) and energy-weighting mode, respectively. The results of the modeled EI system are plotted as circles. (a) 10-cm phantom; (b) 15-cm phantom; (c) 20-cm phantom; (d) 25-cm phantom

LightSpeed VCT using two FDA specified phantoms are illustrated in Fig. 9 (b). It is noted that since CTDI_w were measured in air rather than in PMMA, the simulation data shown in Fig. 9 (b) are the results after conversion by multiplying $(\mu_{en}/\rho)_{\text{PMMA}}^{\text{air}}$, the ratio of the mass energy absorption coefficient in air to that of PMMA. It is observed that the simulation results are in good agreement with the measured values with an average difference of 2%.

3.3. Size dependent kVp and mAs

The detectability index d'^2 is calculated by taking 3D MTF and NPS into Eq (1). Fig. 10 and 11 show the detectability index that is normalized by the delivered phantom dose, d'^2/dose , for phantoms of 10-25 cm diameter, as a function of kVp for the adipose and iodinated blood imaging cases, respectively. The optimal kVp for each phantom in each imaging case is chosen such that the value of d'^2/dose is maximized. It is found that in the PCS system using optimal energy weighting (blue solid lines), the optimal kVp for adipose imaging varies from 50 to 120 kVp as phantom size increases, while an optimal kVp of 60 kVp is observed for all the investigated phantoms in iodinated imaging. For both adipose and iodinated imaging cases, the benefit of the use of low tube potentials is

found substantial compared to the conventional 120 kVp for phantoms with diameter below 20 cm. This is because low-energy x-rays generally contain more contrast information than high-energy x-rays. As phantom diameter increases in adipose imaging, low-energy part of incident spectrum is heavily filtered; hence, the optimal kVp is shifted toward 120 kVp. For iodinated imaging, however, due to the sharply increased attenuation just above the iodine k-edge absorption energy (33.2 keV), a low tube potential of 60 kVp is still desirable for larger phantoms.

To quantify the image improvement resulting from the optimal energy-weighting scheme, the results for the PCS system operated in pure photon-counting mode (dashed lines) are also presented in Fig. 10 and 11. In the case of adipose imaging, the improvement is around 8% for phantoms below 20 cm at optimal kVps, while little improvement is observed for larger phantoms due to heavy filtration of low-energy x-rays. In iodinated imaging, the use of the energy-weighting scheme can improve the detectability index by around 25% for all the investigated phantoms at the conventional 120 kVp, but only 5-7% when the incident spectrum is narrowed down to the optimal 60 kVp.

The results of the modeled EI system are plotted as circles in Fig. 10 and 11. The trends of d'^2/dose as a

Figure 11: Dose-normalized detectability index, d^2/dose , as a function of kVp for iodinated imaging. The results of PCS system are plotted as solid and dashed lines for pure photon-counting (p.c.) and energy-weighting mode, respectively. The results of the modeled EI system are plotted as circles. (a) 10-cm phantom; (b) 15-cm phantom; (c) 20-cm phantom; (d) 25-cm phantom

function of kVp for the investigated phantoms are approximately the same as those of the PCS system for both adipose and iodinated imaging cases. The results also indicate that despite the influences of Compton scatter and charge sharing, our PCS system is superior to the EI system over the kVp and phantom diameter range investigated in the study. With the use of the respective optimal kVps, the dose-normalized detectability indexes of the PCS system are 105-130% of those of the EI system for adipose imaging, and 170% for iodinated imaging.

With the use of the optimal kVp for each phantom in each imaging case, the relative mAs that is necessary to achieve the reference detectability index d_{ref}^2 as defined in section 2.2 is shown in Fig.12: adipose imaging (left figure) and iodinated imaging (right figure). It is clearly shown that a substantial mAs reduction can be achieved for small patients in both imaging cases. When decreasing phantom diameter, the relative mAs is dramatically reduced. To reach the reference detectability index in adipose imaging (Fig.12 (a)), the PCS system requires only 9%, 23%, 24%, and 54% of the mAs that is used for adult patients (30-cm in diameter) at 120 kVp, for 10, 15, 20, and 25-cm phantoms, respectively. While in iodinated imaging (Fig.12 (b)), the reference detectability index can be achieved for

the investigated phantoms when 2%, 9%, 37%, and 109% of the adult mAs are used.

4. Discussion and conclusions

A mathematical framework has been introduced to optimize the scanning parameters (kVp and mAs) for a proposed PCS CT system in pediatric imaging. The results presented in the study show that the scanning parameters for pediatric CT should be selected depending on patient sizes and imaging tasks. Low tube potentials were found optimal for pediatric patients in iodinated imaging, and also optimal in adipose imaging for patients with diameter below 20 cm. One concern for the use of low tube potentials for pediatric patients is that the CT numbers of imaging structures (e.g. soft tissue) might differ from those resulting from 120 kVp; radiologists thus have to adapt to a different overall image impression. However, considering the substantial dose reduction that benefits from the use of low kVps, it is worth making such efforts.

The results have also shown that the our PCS system is superior to the modeled EI system over the kVp and phantom diameter range investigated in the study. One factor not included in the EI system modeling is electronic noise,

Figure 12: The relative mAs necessary to achieve a reference detectability index d'_{ref}^2 , when size-dependent optimal kVps are applied in the PCS and EI systems. (a) adipose imaging; (b) iodinated imaging. For each imaging case, the reference detectability index is given by scanning a 30-cm soft-tissue phantom in a typical adult CT protocol, 120 kVp and 200 mAs, using the modeled EI system. The results for the EI system with electronic noise σ_e included are plotted by crosses. The relative mAs values are displayed in logarithmic scale.

Figure 13: Simulated CT images for a high-resolution task with two 1-mm-diameter iodinated blood vessels separated by 1.5 mm: (a) produced by the modeled energy-integrating system at 60 kVp; (b) produced by the proposed photon-counting spectral system using optimal energy weighting at the same exposure condition.

which can be negligible when a typical adult CT procedure is performed. However, it might become an issue if the quantum noise, determined by the number of detected x-rays, is reduced to a level comparable to intrinsic electronic noise. Since low-energy x-rays generally contain greater contrast information than high-energy x-rays, the x-ray fluence on the detector required for low kVps will be less than that for high kVps to achieve same image quality, thus yielding a reduced quantum noise. To see whether the electronic noise affect the modeled EI system operated at low tube potentials, we added one term in Eq (18) to account for the additive noise caused by electronic noise σ_e ,

$$\text{NPS}(u, v) = \frac{1}{\text{FF}} a_x^2 a_z^2 \left[\frac{\eta \lambda_1^2 \lambda_2^2 \bar{\Phi}}{A_s} + \eta \lambda_1 \lambda_2 \bar{\Phi} \right] + a_x a_z \sigma_e^2 \quad (19)$$

with σ_e assumed to be $4800 e^-$, a value measured from a CT prototype scanner by Luhta et al[34]. The resulting relative mAs for the EI system including electronic noise are plotted by crosses in Fig.12. For the phantom imaging with optimal kVps between 50-60 kVp, a further increase in mAs is required for the EI system to compensate for the effect of electronic noise. An alternative way to reduce the effect of electronic noise is to reduce the total number of CT projections with maintained mAs; however, it would result in angular aliasing artifacts[40], which may affect the detection of fine structures in children. In contrast, our PCS system can reject electronic noise with any number of CT projections by setting a minimum threshold above noise floor, making it more attractive for the application of pediatric imaging.

This study was focused on the detection of a given imaging target. Some imaging tasks also require to determine if an imaging target is separated from other structures. For such high-resolution tasks, more gain can be expected from the proposed PCS system. Fig. 13 shows the simulated CT images of two 1-mm-diameter iodinated blood vessels separated by 1.5 mm, produced by the PSC and EI systems. The PCS system is clearly able to separate the vessels whereas the EI system is not. In future work, it might be of interest to investigate more about such high-resolution tasks.

The pileup effect caused by pulse overlapping of interacting x-rays was not included in the simulations, because it has been measured on our detector module [20] that only 1% of input counts are lost when x-ray fluence rate goes up to $300 \text{ M photons s}^{-1} \text{ mm}^{-2}$ (typical fluence rate range for CT is $1\text{-}200 \text{ M photons s}^{-1} \text{ mm}^{-2}$). In addition, motion unsharpness caused by patients was not modeled in the study and this effect has to be evaluated in future clinical study.

Appendix A. Characterization of bowtie filters

The method to measure the bowtie filter geometry is similar to that proposed by McMillan et al[41]. Fig. 13

Figure A.14: Experimental set-up of the characterization of bowtie filters

shows the experimental setup for the measurement of the bowtie geometry. The CT scanner was operated at 120 kVp in the stationary mode with the x-ray tube fixed at 90° position. An RTI dosimeter was placed at the patient table with the sensitive area at the isocenter. The table was moved along the y-axis in steps of 1 cm and at each step the doses to air with and without bowtie filter were measured. Thus, the ratio D_m of the measured dose in air with bowtie filter to that without bowtie filter can be determined as a function of the fan-beam angle θ :

$$D_m(\theta) = \frac{\text{Dose}_{\text{bowtie}}}{\text{Dose}_{\text{no bowtie}}} \quad (\text{A.1})$$

With known incident spectrum $\Phi(E)$, the theoretical ratio D_t of the dose in air with Al of thickness d_{Al} to that without Al can be calculated:

$$\begin{aligned} D_t(\theta) &= \frac{\text{Dose}_{\text{Al}}}{\text{Dose}_{\text{no Al}}} \\ &= \frac{\int \left(\frac{\mu_{\text{en}}}{\mu} \right)_{\text{air}} \Phi(E) E e^{-\mu_{\text{Al}} d_{\text{Al}}} (1 - e^{-\mu_{\text{air}} d_{\text{air}}}) dE}{\int \left(\frac{\mu_{\text{en}}}{\mu} \right)_{\text{air}} \Phi(E) E (1 - e^{-\mu_{\text{air}} d_{\text{air}}}) dE} \end{aligned} \quad (\text{A.2})$$

where μ is the energy dependent attenuation coefficient, μ_{en} is the energy absorption coefficient, and d_{air} is the air thickness in the dosimeter.

The equivalent Al thickness at each step was determined such that the theoretical dose ratio D_t is equal to the measured dose ratio D_m . Since the bowtie filter is symmetrical along the x-axis, only one half-bowtie filter was measured and the whole bowtie geometry was then obtained by mirroring the results of the measurement.

Appendix B. Derivation of covariance matrix K

For incident x-rays with original energy E , the resulting counts of energy bin i is the sum of primary counts P_i , and double counts S_i that are caused by Compton scat-

tering and charge sharing:

$$g_i = P_i + S_i. \quad (\text{B.1})$$

The variance of bin i , K_{ii} , is then calculated as:

$$\text{Var}[g_i] = \text{Var}[P_i] + \text{Var}[S_i] + 2 \times \text{Cov}[P_i, S_i]. \quad (\text{B.2})$$

Let $\alpha_i(E)$ be the probability of a primary interacting x-ray with original energy E being single registered by bin i , and $\beta_{ij}(E)$ the probability that a primary x-ray produces one count in bin i in its original pixel and one more count in bin j in another pixel. If N energy bins are applied in a photon-counting spectral system, it is expected to be in total $N(N+1)$ possibilities of how an interacting x-ray is registered. Given n primary interacting x-rays, the random output of $N(N+1)$ possibilities follows multinomial distribution. We denote X_i as the output of single registration in bin i and Y_{ij} the output of double-counted events with probability β_{ij} . Then, the random variables P_i and S_i can be expressed as:

$$P_i = X_i + \sum_{k=1}^N Y_{i,k} \quad \text{and} \quad S_i = \sum_{k=1}^N Y_{k,i}. \quad (\text{B.3})$$

By applying the law of total variance, the variance of X_i is calculated as:

$$\begin{aligned} \text{Var}[X_i] &= E[\text{Var}[X_i|n]] + \text{Var}[E[X_i|n]] \\ &= \alpha_i(1 - \alpha_i)E[n] + \alpha_i^2 \text{Var}[n]. \end{aligned} \quad (\text{B.4})$$

Since the number of primary interacting x-rays follows Poisson distribution with the expected value of $\bar{n}$ (i.e., $n \in Po(\bar{n})$), $\text{Var}[X_i]$ is equal to $\alpha_i \bar{n}$. Similarly, the variance of Y_{ij} can be proved to be $\beta_{ij} \bar{n}$. Using the law of total covariance, we show that the outputs of $N(N+1)$ x-ray registration possibilities are independent of each other:

$$\begin{aligned} \text{Cov}(X_i, X_j) &= E[\text{Cov}(X_i, X_j|n)] + \text{Cov}[E(X_i|n), E(X_j|n)] \\ &= -\alpha_i \alpha_j E[n] + \alpha_i \alpha_j \text{Var}[n] \\ &= 0 \quad \text{if } i \neq j. \end{aligned} \quad (\text{B.5})$$

Similarly, $\text{Cov}(X_i, Y_{ij}) = 0$ and $\text{Cov}(Y_{ij}, Y_{kl}) = 0$. Therefore, the variance of P_i and S_i is determined by

$$\begin{aligned} \text{Var}[P_i] &= \text{Var}[X_i + \sum_{k=1}^N Y_{i,k}] = \bar{n}(\alpha_i + \sum_{k=1}^N \beta_{ik}) \\ \text{Var}[S_i] &= \text{Var}[\sum_{k=1}^N Y_{k,i}] = \bar{n} \sum_{k=1}^N \beta_{ki}, \end{aligned} \quad (\text{B.6})$$

and the covariance of P_i and S_i is given:

$$\text{Cov}[P_i, S_i] = \text{Cov}[Y_{i,i}, Y_{i,i}] = \beta_{ii} \bar{n}. \quad (\text{B.7})$$

By substituting Eq B.6 and Eq B.7 into Eq B.2, the

variance of energy bin i for incident energy E is given:

$$\text{Var}[g_i] = \bar{n}[\alpha_i(E) + \sum_{k=1}^N [\beta_{ik}(E) + \beta_{ki}(E)] + 2\beta_{ii}(E)]. \quad (\text{B.8})$$

The covariance of bin i and j , K_{ij} , can also be obtained in a similar way:

$$\text{Cov}[g_i, g_j] = \text{Cov}[Y_{ij} + Y_{ji}, Y_{ij} + Y_{ji}] = [\beta_{ij}(E) + \beta_{ji}(E)] \bar{n} \quad (\text{B.9})$$

For broad incident spectrum $\Phi(E)$ (unit: ph/mm²), the value of K_{ij} is a linear combination of monochromatic results:

$$K_{ij}^m = \begin{cases} \int a_x a_z \Phi^m(E) D(E) [\alpha_i + \sum_{k=1}^N [\beta_{ik} + \beta_{ki}] + 2\beta_{ii}] dE & i = j \\ \int a_x a_z \Phi^m(E) D(E) [\beta_{ij} + \beta_{ji}] dE & i \neq j, \end{cases} \quad (\text{B.10})$$

where $D(E)$ is the detection efficiency of the investigated detector and $\Phi^m(E)$ is the spectrum incident on the detector under the hypothesis h_m .

References

- [1] D. J. Brenner, E. J. Hall, Computed tomography: an increasing source of radiation exposure, *New England Journal of Medicine* 357 (22) (2007) 2277–2284.
- [2] A. Shenoy-Bhangle, K. Nimkin, M. Gee, et al., Pediatric imaging: current and emerging techniques, *Journal of postgraduate medicine* 56 (2) (2010) 98.
- [3] V. Beir, Health risks from exposure to low levels of ionizing radiation, BEIR VII phase 2.
- [4] L. Yu, X. Liu, S. Leng, J. M. Kofler, J. C. Ramirez-Giraldo, M. Qu, J. Christner, J. G. Fletcher, C. H. McCollough, Radiation dose reduction in computed tomography: techniques and future perspective, *Imaging in medicine* 1 (1) (2009) 65–84.
- [5] N. research council. Committee to assess health risks from exposure to low levels of ionizing radiation, *Health Risks from Exposure to Low Levels of Ionizing Radiation: Beir VII Phase II*, National Academic Press, 2006.
- [6] P. Hall, H.-O. Adami, D. Trichopoulos, N. L. Pedersen, P. Lagiou, A. Ekblom, M. Ingvar, M. Lundell, F. Granath, Effect of low doses of ionising radiation in infancy on cognitive function in adulthood: Swedish population based cohort study, *Bmj* 328 (7430) (2004) 19.
- [7] W. Huda, K. A. Lieberman, J. Chang, M. L. Roskopf, Patient size and x-ray technique factors in head computed tomography examinations. i. radiation doses, *Medical physics* 31 (3) (2004) 588–594.
- [8] D. P. Frush, Strategies of dose reduction, *Pediatric radiology* 32 (4) (2002) 293–297.
- [9] R. K. Zeman, R. L. Baron, R. B. Jeffrey Jr, J. Klein, M. J. Siegel, P. M. Silverman, Helical body ct: evolution of scanning protocols., *AJR. American journal of roentgenology* 170 (6) (1998) 1427–1438.
- [10] L. F. Donnelly, K. H. Emery, A. S. Brody, T. Laor, V. M. Gylys-Morin, C. G. Anton, S. R. Thomas, D. P. Frush, Minimizing radiation dose for pediatric body applications of single-detector helical ct: strategies at a large children's hospital, *American Journal of Roentgenology* 176 (2) (2001) 303–306.
- [11] J. M. Boone, E. M. Geraghty, J. A. Seibert, S. L. Wootton-Gorges, Dose reduction in pediatric ct: A rational approach 1, *Radiology* 228 (2) (2003) 352–360.

- [12] W. A. Kalender, P. Deak, M. Kellermeier, M. van Straten, S. V. Vollmar, Application-and patient size-dependent optimization of x-ray spectra for ct, *Medical physics* 36 (3) (2009) 993–1007.
- [13] D. D. Cody, D. M. Moxley, K. T. Krugh, J. C. O'Daniel, L. K. Wagner, F. Eftekhari, Strategies for formulating appropriate mdct techniques when imaging the chest, abdomen, and pelvis in pediatric patients, *American Journal of Roentgenology* 182 (4) (2004) 849–859.
- [14] M. J. Siegel, B. Schmidt, D. Bradley, C. Suess, C. Hildebolt, Radiation dose and image quality in pediatric ct: Effect of technical factors and phantom size and shape 1, *Radiology* 233 (2) (2004) 515–522.
- [15] Y. Funama, K. Awai, Y. Nakayama, K. Kakei, N. Nagasue, M. Shimamura, N. Sato, S. Sultana, S. Morishita, Y. Yamashita, Radiation dose reduction without degradation of low-contrast detectability at abdominal multislice ct with a low-tube voltage technique: Phantom study 1, *Radiology* 237 (3) (2005) 905–910.
- [16] D. P. Frush, J. R. Herlong, Pediatric thoracic ct angiography, *Pediatric radiology* 35 (1) (2005) 11–25.
- [17] D. J. Brenner, Estimating cancer risks from pediatric ct: going from the qualitative to the quantitative, *Pediatric radiology* 32 (4) (2002) 228–231.
- [18] J. Ramirez-Giraldo, A. Primak, K. Grant, B. Schmidt, M. Fuld, Radiation dose optimization technologies in multidetector computed tomography: A review, *MEDICAL PHYSICS* 2 (2).
- [19] C. Xu, H. Chen, M. Persson, S. Karlsson, M. Danielsson, C. Svensson, H. Bornefalk, Energy resolution of a segmented silicon strip detector for photon-counting spectral ct, *Nuclear Instruments and Methods in Physics Research Section A: Accelerators, Spectrometers, Detectors and Associated Equipment* 715 (2013) 11–17.
- [20] C. Xu, M. Persson, H. Chen, S. Karlsson, M. Danielsson, C. Svensson, H. Bornefalk, Evaluation of a second-generation ultra-fast energy-resolved asic for photon-counting spectral ct, *IEEE Transactions on Nuclear Science* 60 (1) (2013) 437–445.
- [21] M. Persson, B. Huber, S. Karlsson, X. Liu, H. Chen, C. Xu, M. Yveborg, H. Bornefalk, M. Danielsson, Energy-resolved ct imaging with a photon-counting silicon-strip detector, in: *SPIE Medical Imaging, International Society for Optics and Photonics*, 2014, pp. 90333L–90333L.
- [22] H. Bornefalk, M. Persson, C. Xu, S. Karlsson, C. Svensson, M. Danielsson, Effect of temperature variation on the energy response of a photon counting silicon ct detector, *Nuclear Science, IEEE Transactions on* 60 (2) (2013) 1442–1449.
- [23] X. Liu, H. Bornefalk, H. Chen, M. Danielsson, S. Karlsson, M. Persson, C. Xu, B. Huber, A silicon-strip detector for photon-counting spectral ct: energy resolution from 40 kev to 120 kev, *Nuclear Science, IEEE Transactions on* 61 (3) (2014) 1099–1105.
- [24] P. M. Shikhaliev, Projection x-ray imaging with photon energy weighting: experimental evaluation with a prototype detector, *Physics in medicine and biology* 54 (16) (2009) 4971.
- [25] T. G. Schmidt, Ct energy weighting in the presence of scatter and limited energy resolution, *Medical physics* 37 (3) (2010) 1056–1067.
- [26] Technical Reference Manual: LightSpeed VCT, General Electric Company, 2008.
- [27] B. Zhao, J. Zhou, Y.-H. Hu, T. Mertelmeier, J. Ludwig, W. Zhao, Experimental validation of a three-dimensional linear system model for breast tomosynthesis, *Medical physics* 36 (1) (2009) 240–251.
- [28] M. Yveborg, M. Danielsson, H. Bornefalk, Performance evaluation of a sub-millimetre spectrally resolved ct system on high-and low-frequency imaging tasks: a simulation, *Physics in medicine and biology* 57 (8) (2012) 2373.
- [29] S. Richard, E. Samei, Quantitative imaging in breast tomosynthesis and ct: Comparison of detection and estimation task performance, *Medical physics* 37 (6) (2010) 2627–2637.
- [30] D. J. Tward, J. H. Siewerdsen, Cascaded systems analysis of the 3d noise transfer characteristics of flat-panel cone-beam ct, *Medical physics* 35 (12) (2008) 5510–5529.
- [31] J. Beutel, J. M. Fitzpatrick, S. C. Horii, Y. Kim, H. L. Kundel, M. Sonka, R. L. Van Metter, *Handbook of Medical Imaging, Volume 3. Display and PACS*, Washington, DC: SPIE Press, 2002.
- [32] H. H. Barrett, K. J. Myers, *Foundations of image science, Foundations of Image Science*, by Harrison H. Barrett, Kyle J. Myers, pp. 1584. ISBN 0-471-15300-1. Wiley-VCH, October 2003. 1.
- [33] J. H. Siewerdsen, D. A. Jaffray, Optimization of x-ray imaging geometry (with specific application to flat-panel cone-beam computed tomography), *Medical physics* 27 (8) (2000) 1903–1914.
- [34] R. Luhta, M. Chappo, B. Harwood, R. Mattson, D. Salk, C. Vrettos, A new 2d-tiled detector for multislice ct, in: *Medical Imaging, International Society for Optics and Photonics*, 2006, pp. 61420U–61420U.
- [35] H. Bornefalk, M. Danielsson, Photon-counting spectral computed tomography using silicon strip detectors: a feasibility study, *Physics in medicine and biology* 55 (7) (2010) 1999.
- [36] C. Xu, H. Chen, M. Persson, S. Karlsson, M. Danielsson, C. Svensson, H. Bornefalk, Energy resolution of a segmented silicon strip detector for photon-counting spectral ct, *Nuclear Instruments and Methods in Physics Research A* 715 (2013) 11–17.
- [37] J. A. Bauhs, T. J. Vrieze, A. N. Primak, M. R. Bruesewitz, C. H. McCollough, Ct dosimetry: Comparison of measurement techniques and devices 1, *Radiographics* 28 (1) (2008) 245–253.
- [38] I. E. Commission, et al., Medical electrical equipment. part 2-44, particular requirements for the safety of x-ray equipment for computed tomography, IEC standard (2002) 60601–2.
- [39] U. S. Food, D. A. (FDA), Diagnostic x-ray systems and their major components, Code of Federal Regulations 21 CFR (1020.33).
- [40] J. F. Barrett, N. Keat, Artifacts in ct: recognition and avoidance 1, *Radiographics* 24 (6) (2004) 1679–1691.
- [41] K. McMillan, M. McNitt-Gray, D. Ruan, Development and validation of a measurement-based source model for kilovoltage cone-beam ct monte carlo dosimetry simulations, *Medical physics* 40 (11) (2013) 111907.