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Assessment of X-Ray Crosstalk in a Computed Tomography Scanner with Small Detector Elements Using Monte Carlo Method

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ABSTRACT

Introduction: Crosstalk is a leakage of X-ray or light produced in a matrix of X-ray detectors or array of photodiodes in one element to other elements affecting on image contrast and spatial resolution. In this study, we assessed X-ray crosstalk in a computed tomography (CT) scanner with small detector elements to estimate the effect of various parameters such as X-ray tube voltage, detector element sizes, scintillator material, impurities in the scintillator material, and the material of detector separators on X-ray crosstalk.

Materials and Methods: This study was performed using Monte Carlo simulation. In the first step, X-ray tube and its energy spectrum at the energies of 80, 100, 120, and 140 keV were simulated and validated by using SpekCalc and t-test. Then, other important parts of CT scanner, namely filters, detectors, and grids were simulated. X-ray crosstalk between CT detectors was calculated in air and in the presence of water phantom (as a simulator of human body) to compare the effect of scattered photons. Finally, the influence of some important parameters on X-ray crosstalk was evaluated.

Results: In CT scanner with small elements, when using phantom, crosstalk increases by 16-50%. Using the lowest possible energies of X-ray, decreases the crosstalk up to 43% of its initial amount. Furthermore coating a 10 or 20 µm layer of tungsten or lead on the detector separators, decreases the X-ray crosstalk significantly.

Conclusion: Choosing the proper high voltage, detectors' material and its dimensions, scintillator impurities and septa material can decrease X-ray crosstalk.

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Introduction

Computed tomography (CT) can provide high-spatial-resolution images of objects and the human anatomy. Increased spatial resolution requires detectors with a large number of elements with small dimensions. In multi-slice CT technology, detectors have been used to increase the scanning speed, and sub-millimeter slice can be used to improve spatial resolution. By curtailing the dimensions of the scintillator layer elements, the dimensions of photodiode layer elements decrease and the probability of crosstalk increases [1, 2].

Crosstalk is the leakage of X-ray and/or light produced from the irradiated center pixel to the non-radiated neighbor pixels due to the leakage of X-ray and/or light produced in scintillators. Crosstalk is one of the most important parameters in the detectors for X-ray imaging equipment, and it impacts on the output of detectors and image quality [1, 3]. In scintillator detectors, there are four kinds of crosstalk sources, which one of them is electrical and the others represent different types of optical crosstalk [4].

Considering the importance of this issue, various research teams in several studies have evaluated crosstalk types and their reduction methods in variable resolution X-ray (VRX) CT scans. The calculation method of the optical crosstalk in a 9×9 array of detector elements was suggested by Ikhlef and Thrivikraman by using the DETECT code in 2004 [5]. To calculate X-ray crosstalk, they simulated the deposited energy for the exposed to radiation pixel and in the neighboring pixels, and the energy deposited in the non-exposed pixel showed the scattered X-ray from one pixel to the other. Electrical crosstalk in high-density photodiode arrays for X-ray imaging techniques was evaluated by Ji et al. in 2009 [1], and they investigated the electrical crosstalk in front-illuminated photodiode arrays with different guard ring designs in another study in 2009 [6]. Inter-

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cell X-ray crosstalk in VRX CT scanner was simulated using the Monte Carlo method as a function of the detector field of view (FOV) by Melnyk and DiBianca in 2003. They reported that the inter-cell X-ray crosstalk was maximum at the 34-36 cm FOV [7]. Further, Arabi et al. in 2011 proposed a new cell arrangement in VRX CT scanner to decrease inter-cell X-ray crosstalk in 24-38 cm FOV [8].

One type of optical crosstalk is X-ray crosstalk that is created due to X-rays scattering from one scintillator element to its neighbors [4]. In previous studies, the X-ray crosstalk in multi-slice CT scanners was calculated without phantom modeling. Previous modeling results have not been reported quantitatively, and the effects of various parameters on X-ray crosstalk have not been studied. In this study, different parts of a 64-slice CT scanner such as X-ray tube, filters, detectors, and grids were simulated using Monte Carlo N-particle radiation transport computer code (MCNP5). By placing water phantom in the center of gantry, calculation method of X-ray crosstalk in 64-slice CT detectors was modeled and its results were compared with the amount of X-ray crosstalk in the case where the phantom was not used. Then, the effects of various parameters such as X-ray energy spectra, dimensions of detector elements, composition of the scintillating material, and impurities in the scintillator and separator materials on X-ray crosstalk were determined.

Materials and Methods

Simulation of X-ray tube and its energy spectrum

In this study, the MCNP5 code (Los Alamos National Laboratory, University of California. USA) was utilized to assess X-ray crosstalk in a 64-slice CT scanner with small elements. A GE 64-slice LightSpeed VCT CT scanner (General Electric Healthcare Technologies, Waukesha, WI, USA) was used in this study. In the first step, the X-ray tube of the scanner was simulated. The X-ray tube has large focal spot size (0.9×0.9 mm), which is made of tungsten, and the target (anode) vertical angle is 7 degrees according to the research by Akbarzadeh et al. [9]. The X-ray spectrum simulation procedure was started by defining an electron source as an emitting point of electrons with energy of E and angle Ψ toward the focal spot [9, 10]. E was considered as 80, 100, 120, and 140 keV in the simulations because of the scanner specifications [11]. Additionally, the maximum mA of the scanner (1500 mA) was considered in fixed current mode. The X-ray beam width was considered equal to 10 mm [12]. Electrons, after hitting the anode, were tracked by Monte Carlo simulation until they were absorbed or emerged from it.

The production of bremsstrahlung and characteristic photons was considered during electron transport within the target. The X-rays passed through the Be window with a thickness of 1 mm [10]. After that, the X-ray spectrum passed through the inherent and additional filters, and then photon flux was measured at the simulation output. The scanner has inherent filtration of 3.25 mm Al and added filtration of 0.1 mm Cu [13]. By using the F2 tally, photon flux at the surface was measured as photons/cm^2. Then, the tally was normalized to the number of starting particles. Figure 1 shows the geometry of X-ray tube used in Monte Carlo simulation of X-ray spectra in this study.

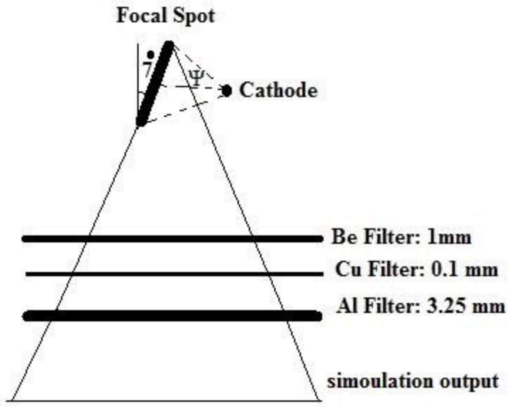

Figure 1. The geometry and components of X-ray tube used in this study to calculate the X-ray spectra in various kVps

Simulation of phantom

To assess the effect of phantom on X-ray crosstalk, we used a water-filled cylindrical phantom made of Perspex with an external diameter of 215 mm, a wall thickness of 6 mm, and a height of 700 mm [9].

Simulation of detectors and grids

The LightSpeed 64 slice CT scanner has 58368 individual elements arranged in 57 modules with 64 rows of 0.625 mm thickness in the z-direction and sixteen 0.55 mm channels in the x-direction. The thickness of detector elements is 3 mm in the y-direction. The scanner is equipped with Highlight ($Y_2Gd_2O_3$:Eu) ceramic scintillators composed of gadolinium and yttrium and an europium activator [9, 14, 15]. The one-dimensional anti-scatter grid with a grid ratio of 10 was placed on the detector module. The grid has a simple geometric design, which has open interspace regions and alternating X-ray absorbing tungsten septa with a thickness of 0.2 mm. We employed lattice geometry to define the geometry of the detectors and grids [16]. Figure 2 shows one element (a) and a 5×5 array of detectors (b) that were simulated. In the GE LightSpeed 64 slice CT scanners, the block detectors have the array of 16×64 of elements, but to evaluate X-ray crosstalk, a matrix array of 5×5 should be proper to reduce the computational time of calculations. After simulating different parts of the scanner, the detector module was adjusted at a distance of 95 cm from the focal spot and 41 cm from the gantry center according to the study by Akbarzadeh et al. [9].

Calculation of x-ray crosstalk in CT detectors

Calculation of X-ray crosstalk in CT detectors without using the phantom

By considering the 5×5 array of detector module, the light-sensitive area of the central element was radiated by X-ray photons with a 140 kVp energy spectrum. The energy deposited in the neighboring elements show X-ray scattering within the scintillator from the central element to its neighboring elements according to the study of Ikhlef and Thrivikraman [5].

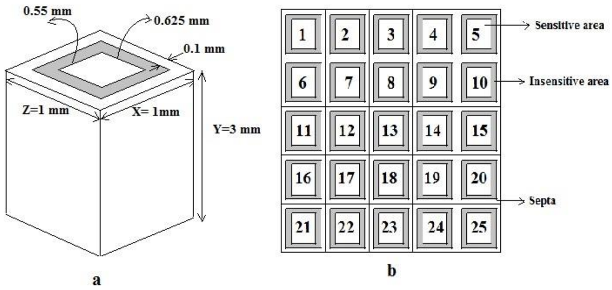

Figure 2. (a) One element and (b) a 5×5 array of detectors in GE LightSpeed 64 slice computed tomography scanner, its dimensions, and arrangement

Calculation of the X-ray crosstalk in CT detectors in the phantom model

By considering the water phantom at the center of gantry, part of the deposited energy in the neighboring elements could have resulted from photons that were scattered in the phantom; thus, X-ray crosstalk could not be calculated by measuring the energy deposited in neighboring elements. To calculate X-ray crosstalk, scattered radiations must be separated from the primary photons. For this purpose, a virtual plane was considered after the phantom and the surface source method implemented in the MCNP5 code was used. By using surface source write (SSW) card, the history of each photon passing through this plane was recorded. In the next step, by using surface source read (SSR) card and its collision flag (COL), the primary components were calculated by considering only photons that had no interaction before hitting the virtual plane. Then, the energy deposited in the central element and its adjacent elements were measured by using the F6 tally.

Figure 3 presents a typical diagram calculation of the amount of X-ray crosstalk in the case of using phantom mode (a) and not using the phantom (b). The number of photon histories (NPS) in each run of the Monte Carlo code was 10^9.

Evaluation of the influence of some important parameters on X-ray crosstalk

The detector modulation transfer function (MTF) was applied to characterize the influence of the most important parameters, such as X-ray energy spectra, dimensions of detector elements, composition of the scintillating material, and impurities in the scintillator and separator materials, on X-ray crosstalk.

The quantification of detector MTF was performed by evaluating the corresponding line spread-function (LSF) [8]. To evaluate the LSF, a line source with a 140 kVp energy spectrum irradiated a corresponding column of detectors. Afterwards, a profile was measured perpendicular to this line source. This profile is the acquired data from the central element which is on the mentioned column of detectors and 7 neighbor pixels in left and right. The LSF was normalized to have unity area, and then, the modulation Fourier transform of LSF was computed to achieve the MTF.

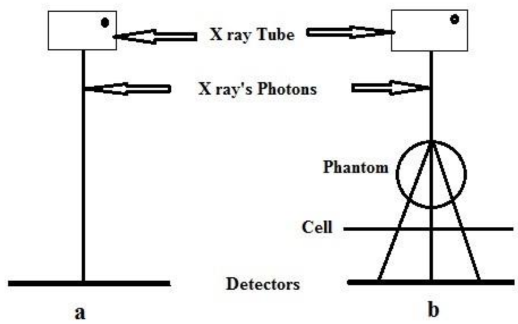

Figure 3. The used settings to calculate X-ray crosstalk in the absence of phantom (a) and in phantom model (b)

Results

Validation of the simulated X-ray spectra

Figure 4 exhibits the simulated X-ray spectra for 80, 100, 120, and 140 kVp tube voltages using the Monte Carlo method. Comparison of the Monte Carlo simulated with SpekCalc spectra for tube voltage of 140 kVp has been shown in Figure 5. SpekCalc is executable to calculate the X-ray emission spectra from tungsten anodes such as those used in diagnostic radiology [17]. Quantitative evaluation of the differences between Monte Carlo simulation and SpekCalc spectra was performed using student's t-test. The t-test showed no statistically significant differences between the Monte Carlo method and SpekCalc spectra at the 95% confidence level in this study.

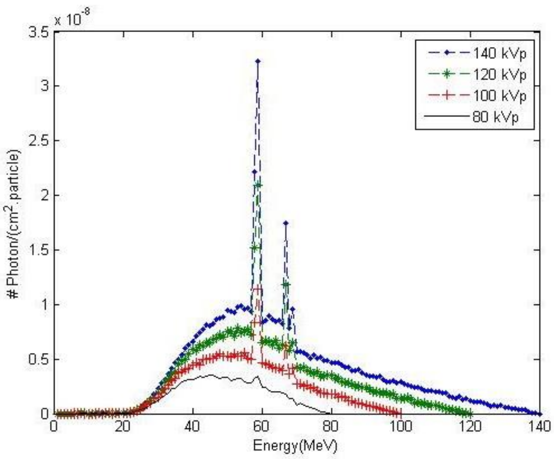

Figure 4. The simulated X-ray spectra (140, 120, 100, and 80 kVp) in this study by using the Monte Carlo method

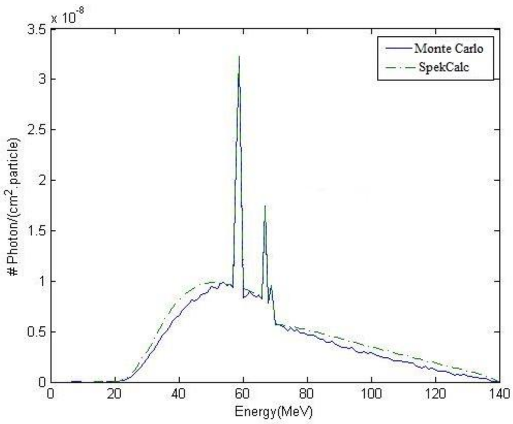

Figure 5. Comparison of the Monte Carlo simulated with SpekCalc spectrum for tube voltage of 140 kVp

The calculated X-ray crosstalk in CT scanner detectors

The percentages of calculated X-ray crosstalk in GE LightSpeed 64 slice CT scanner by using the phantom and in the case without using the phantom have been reported in Error! Reference source not found.. As can be noted in Figure 2-b, element 13 is the central element. Elements with the numbers of 8, 12, 14, and 18 (3, 11, 15, and 23) are the first (second) horizontal and vertical neighbors, and elements 7, 9, 17, and 19 (1, 5, 21, and 25) are the first (second) diagonal neighbors.

By considering the results in Error! Reference source not found., in the first and second horizontal and vertical (diagonal) neighbors of the central element, the amount of primary X-ray crosstalk when using phantom mode was higher than the X-ray crosstalk when not using phantom mode. In fact, by the collision of X-ray photons to the phantom, low-energy photons are removed, and in comparison to the case that phantom is not in the direction of X-ray photons, the average amount of photon energies that reach the detector is increased. Therefore, the possibility of Compton scattering of X-ray photons that reached the detector increases.

The effects of various parameters on X-ray crosstalk

The effect of different energy spectra on X-ray crosstalk

By applying a high voltage to the X-ray tube, the effect of 140, 120, 100 and 80 kVp energy spectra on X-ray crosstalk was investigated. As Figure 6 shows, for MTF of 0.98, spatial resolutions were calculated to be 4.2, 4.5, 5, and 6 cycles/mm, respectively (MTF of 0.98 has been selected based on the obtained MTFs so that minor variations in the resolution can be calculated). In fact, by increasing X-ray energy, the probability of Compton scattering increases, hence elevated X-ray crosstalk.

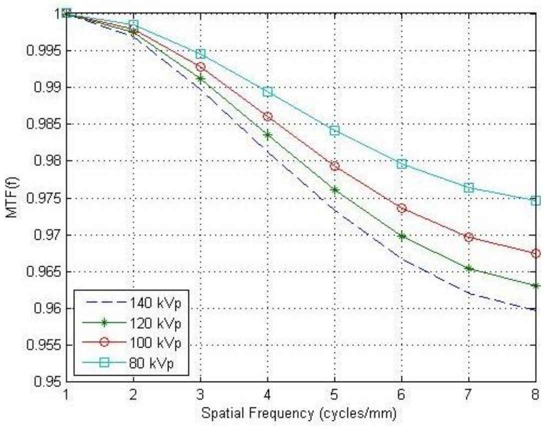

Figure 6. Plot of the detector modulation transfer function in various energy spectra (140, 120, 100, and 80 kVp)

Table 1. The percentage of calculated X-ray crosstalk in a GE 64-slice Light Speed VCT scanner with and without using phantom

Simulation mode	Crosstalk in the first horizontal and vertical neighbors	Crosstalk in the first diagonal neighbors	Crosstalk in the second horizontal and vertical neighbors	Crosstalk in the second diagonal neighbors	The calculated error by using Monte Carlo method
Without using phantom	0.44	0.17	0.019	0.006	0.01
With using phantom	0.51	0.2	0.027	0.009	0.08

The effect of detector elements' dimensions on X-ray crosstalk

To quantitatively investigate the effect of scintillator elements' dimensions on X-ray crosstalk, the dimensions of scintillator elements were changed in the three directions of X, Y, and Z.

Figure 7 demonstrates that for MTF of 0.98, by changing the size of scintillator elements in the X direction (Figure 2-a) from 0.55 mm to 0.6 and 0.5 mm, the spatial resolution changed from 4.2 cycles/mm to 4.4 and 4 cycles/mm, respectively.

Figure 8 presents that for MTF of 0.98, by altering the thickness of scintillator elements in the Y direction (Figure 2-a) from 3 mm to 2.5 and 3.5 mm, the spatial resolution changed from 4.2 cycles/mm to 4.4 and 4 cycles/mm, respectively.

According to Figure 9, for MTF of 0.98, by changing the slice thickness of scintillator elements in the Z direction (Figure 2-a) from 0.625 mm to 0.6 and 0.5 mm, the spatial resolution changed from 4.2 cycles/mm to 4.1 and 3.7 cycles/mm, respectively.

Changing the size of the detector elements indicated that by curtailing the surface of scintillator elements and expanding their thickness, the amount of X-ray crosstalk increases. In addition, the impact of changing detector elements' surface (X, Z) on the amount of X-ray crosstalk is higher than that of the thickness of detector elements (Y).

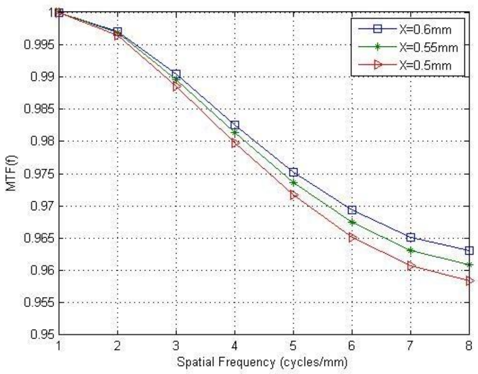

Figure 7. Plot of the detector modulation transfer function in various scintillator element sizes (0.6, 0.55, and 0.5 mm) in the X direction

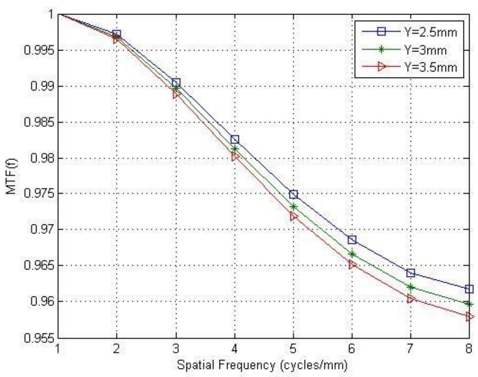

Figure 8. Plot of the detector modulation transfer function in various scintillator element thicknesses (2.5, 3, and 3.5 mm) in the Y direction

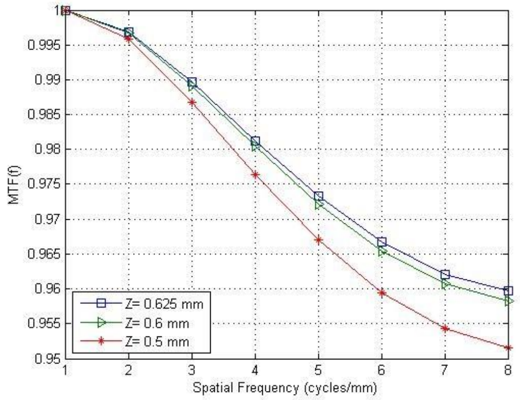

Figure 9. Plot of the detector modulation transfer function in various slice thicknesses of scintillator elements (0.5, 0.6, and 0.625 mm) in the Z direction

The effect of scintillation material on X-ray crosstalk

To ascertain the effect of scintillator material on X-ray crosstalk, the material of GE scintillator was changed from $Y_2 Gd_2 O_3$: Eu to $CdWO_4$, GOS (material of Toshiba and Philips scintillators) and BGO (material of Siemens scintillator). As Figure 10 shows, for MTF of 0.98, spatial resolution was changed from 4.2 cycles/mm to 4.5, 3.7 and 3.1 cycles/mm, respectively. In fact, by increasing the atomic number and density of detector constituents, photoelectric phenomenon is more likely to occur than Compton scattering, and X-ray crosstalk is decreased.

The effect of scintillator impurities on X-ray crosstalk

By removing the europium impurities from the GE scintillator material, the effect of GE scintillator impurities on X-ray crosstalk was investigated. As Figure 11 illustrates, for MTF of 0.98, the spatial resolutions were calculated to be 4.2 and 4 cycles/mm using the europium impurities and in their absence, respectively.

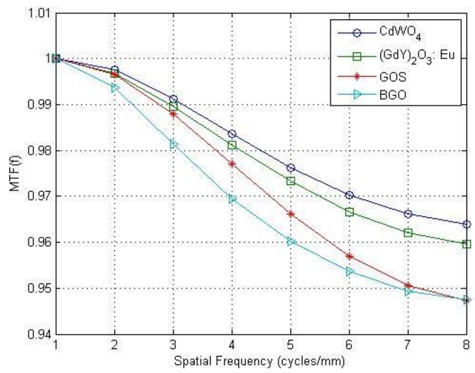

Figure 10. Plot of the detector modulation transfer function in various scintillator materials (CdWO4, Y2Gd2O3: Eu, GOS, BGO)

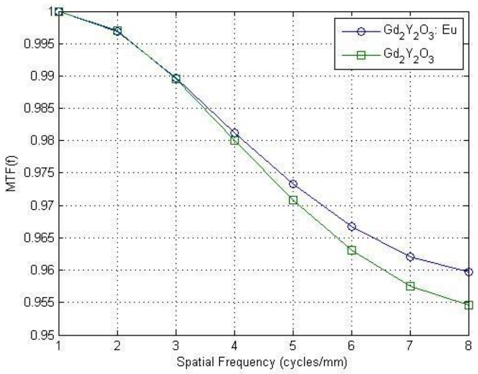

Figure 11. Plot of the detector modulation transfer function without europium impurities in GE scintillator material ($Y_2Gd_2O_3$) and using GE scintillator material ($Y_2Gd_2O_3$: Eu)

The effect of septa material in CT scan detector on X-ray crosstalk

In GE detectors, septa may not remove scattered X-rays and cross-talk completely. Therefore, a layer of tungsten or lead with a thickness of 10 µm, and then 20 µm, was added to the separators in GE detectors, and its effect on X-ray crosstalk was studied. As Figure 12 shows, for MTF of 0.985 when the septa is made of, pure polymer, pure polymer and layers of tungsten or lead with a thickness of 10 (20) µm, the spatial resolutions were calculated and they are equal to 3.5, 5, and 4.5 cycles/mm (7 and 5.6 cycles/mm), respectively.

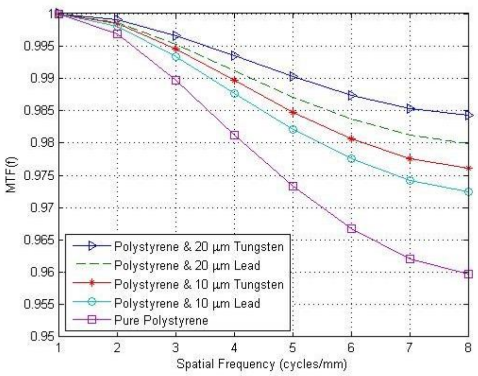

Figure 12. Plot of the detector modulation transfer function in various separator materials (pure polymer, an added layer of tungsten or lead with thickness of 10 µm or 20 µm to the separators)

Discussion

To calculate X-ray crosstalk in CT scanners including small detector sizes, we utilized a 64-slice CT scanner and Monte Carlo simulation. To consider the effect of scattered photons, a phantom was employed. The results indicated that the amount of X-ray crosstalk in the first neighborhood of the central element using the phantom is 20% higher than that in the case without using the phantom, while Ikhlef et al. (2004) did not consider the effect of phantom on the amount of X-ray crosstalk. X-ray crosstalk in the first horizontal and vertical neighbors of the central element is 26 times higher than the X-ray crosstalk in the first diagonal neighbors of the central element. The calculated spatial resolution was 4.2 cycles/mm in the GE LightSpeed VCT 64-slice CT scanner for MTF of 98% and energy spectrum of 140 kVp. When using 120, 100, and 80 kVp energy spectra instead of 140 kVp, spatial resolution increased by 7%, 19%, and 43%, respectively. In the energy spectrum of 140 kVp, for each reduction of 0.05 mm (0.025 mm) in the size of detector elements in the direction of X (Z) axis from 0.6 to 0.5 mm (from 0.625 to 0.5 mm), spatial resolution reduced by 5% (2.5%). By enhancing the detector thickness from 2.5 to 3.5 mm with the interval of 0.5 mm, spatial resolution diminished by 5%.

By changing the detectors' material from $Y_2Gd_2O_3$: Eu (GE company) to $CdWO_4$ (GOS [Toshiba and Philips companies] and BGO [Siemens company]), spatial resolution increased (reduced) by 7% (12% and 26%). By removing the europium impurities from GE detectors' material, spatial resolution reduced by 5%. For MTF of 98.5%, by adding a layer of tungsten (lead) with a thickness of 10 µm to the separators in GE detectors, spatial resolution was enhanced by 43% (29%).

Conclusion

Based on our simulation results, to decrease X-ray crosstalk in CT scan, it is necessary that the voltage of X-ray tube be chosen best by the operator. It means the high voltage should be select as low as possible. The use of detector elements with small surface and high thickness can increase crosstalk in GE 64-slice CT scanner. In addition, the impact of detector elements' surface (X, Z) on the amount of X-ray crosstalk is higher relative to the thickness of detector elements (Y). Given our results, spatial resolution was increased by using $CdWO_4$ detectors instead of $Y_2Gd_2O_3$: Eu detectors. Separators in GE LightSpeed VCT detectors are made of pure polymer. To reduce X-ray crosstalk, we recommend coating separators with a layer of tungsten or tungsten powder.

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