Reducing radiation dose to selected organs by selecting the tube start angle in MDCT helical scans: A Monte Carlo based study

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Purpose: Previous work has demonstrated that there are significant dose variations with a sinusoidal pattern on the peripheral of a CTDI 32 cm phantom or on the surface of an anthropomorphic phantom when helical CT scanning is performed, resulting in the creation of “hot” spots or “cold” spots. The purpose of this work was to perform preliminary investigations into the feasibility of exploiting these variations to reduce dose to selected radiosensitive organs solely by varying the tube start angle in CT scans.

Methods: Radiation dose to several radiosensitive organs (including breasts, thyroid, uterus, gonads, and eye lenses) resulting from MDCT scans were estimated using Monte Carlo simulation methods on voxelized patient models, including GSF’s Baby, Child, and Irene. Dose to fetus was also estimated using four pregnant female models based on CT images of the pregnant patients. Whole-body scans were simulated using 120 kVp, 300 mAs, both 28.8 and 40 mm nominal collimations, and pitch values of 1.5, 1.0, and 0.75 under a wide range of start angles (0° – 340° in 20° increments). The relationship between tube start angle and organ dose was examined for each organ, and the potential dose reduction was calculated.

Results: Some organs exhibit a strong dose variation, depending on the tube start angle. For small peripheral organs (e.g., the eye lenses of the Baby phantom at pitch 1.5 with 40 mm collimation), the minimum dose can be 41% lower than the maximum dose, depending on the tube start angle. In general, larger dose reductions occur for smaller peripheral organs in smaller patients when wider collimation is used. Pitch 1.5 and pitch 0.75 have different mechanisms of dose reduction. For pitch 1.5 scans, the dose is usually lowest when the tube start angle is such that the x-ray tube is posterior to the patient when it passes the longitudinal location of the organ. For pitch 0.75 scans, the dose is lowest when the tube start angle is such that the x-ray tube is anterior to the patient when it passes the longitudinal location of the organ.

Conclusions: Helical MDCT scanning at pitch 1.5 and pitch 0.75 results in “cold spots” and “hot spots” that are created both at surface and in-depth locations within patients. For organs that have a relatively small longitudinal extent, dose can vary considerably with different start angles. While current MDCT systems do not provide the user with the ability to control the tube start angle, these results indicate that in these specific situations (pitch 1.5 or pitch 0.75, small organs and especially small patients), there could be significant dose savings to organs if that functionality would be provided. © 2009 American Association of Physicists in Medicine.

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I. INTRODUCTION

Radiation dose to patients continues to be a significant concern to medical physicists and to the broader medical community as well. Radiation dose from CT exams has been identified as the largest source of medical radiation exposure.¹ Specifically for CT scans, various approaches have been developed to reduce radiation dose, including tube

current modulation (TCM),² adjusting/lowering mAs for patient size,^{3,4} and lowering kVp (especially for studies using iodinated contrast⁵), while maintaining diagnostic image quality.

One commonly used metric to evaluate risk from radiation is the effective dose, calculated from a weighted average of the absorbed dose to individual radiosensitive organs as

FIG. 1. The dose distribution along the z axis on the surface of a CTDI 32 phantom undergoing a CT scan with nominal beam width of 28.8 mm and pitch 1.5. The locations of x-ray beam on, x-ray beam off, start image data, and end image data are also shown. Double headed arrows indicate regions of overscan.

defined by ICRP.⁶ While effective dose is very useful, it has been suggested that a better method to estimate risk would be to estimate the radiation dose to specific radiosensitive organs, which can be used in biological models to estimate the probability of radiation induced carcinogenesis and genetic effects.^{6–10}

Previous work has shown that the surface dose distribution resulting from helical scanning of either a standard dosimetry phantom (32 cm CTDI phantom), or an anthropomorphic phantom, is periodic.^{11,12} It was also shown¹¹ that the period is influenced by the actual radiation beam width and table movement. This work also demonstrated that the magnitude of variation could be large, with maximum values greater than twice the minimum values under some conditions (e.g., pitch=1.5). This magnitude of variation has been shown in both homogeneous cylindrical phantoms as well as heterogeneous anthropomorphic phantoms.

Because of the periodic nature of this dose pattern, the effect of the tube start angle (essentially determining a phase shift of the dose distribution) was investigated and it was shown that the tube start angle determined the locations of the high dose and low dose regions. For example, for a scan using pitch 1.5 and nominal beam width of 24×1.2 mm, the simulated longitudinal dose distribution at the surface of a 32 cm diameter CTDI phantom along the z axis is shown in Fig. 1. This figure also describes the region where the x-ray beam turns on and off and the beginning and end of image data collection. The shape of the dose variation usually appears as a sinusoidal function because of the rotation of the CT tube. This creates “cold spots” (locations where the dose is minimum) along the z axis. As the tube start angle changes, the distribution of these cold spots will “move” along the z axis.

This concept is illustrated in Fig. 2, which shows a divergent x-ray beam and the corresponding surface dose distribution along the z axis under two scenarios with different tube start angles. An illustrative patient with a schematic indication of the location of the breasts is shown at the bottom of the figure as well. In the upper scenario, the cold spots occur at the location of the breasts, so the breast would receive lower dose, as opposed to the lower scenario, where the breasts would receive high dose.

FIG. 2. Diagram illustrating the tube position and corresponding surface dose distribution along the z axis under two scenarios with different tube start angles at pitch 1.5. An illustrative patient with schematic indication of the location of the breasts is shown at the bottom as well. This figure shows that the dose to the breasts of the patient can be affected by selecting different tube start angles.

Therefore, the purpose of this work was to extend the previous simulation work to investigate the effects these phase shifts, caused by different start angles, on organ doses in patients. While current MDCT systems do not allow the user to control the start angle, this study investigates the feasibility of exploiting this sinusoidal behavior to reduce the dose to a specific organ by varying the tube start angle, especially for a small peripheral organ on small patients. To do this, we simulated CT scans under different conditions (pitch, collimation, etc.) for different patient models (infant, small child, adult female, and pregnant patient) under different start angle conditions to determine the effects on organ dose and to determine how much organ dose reduction was possible.

II. MATERIAL AND METHODS

II.A. Monte Carlo method

The Monte Carlo method has been used to simulate radiation dose deposition from CT in several previous efforts.^{13–20} For this work, a Monte Carlo particle transport package, MCNP eXtended v2.6 (MCNPX), which was developed at Los Alamos National Laboratory, was utilized to perform the simulations. The simulations were operated in photon transport mode with a low-energy cutoff of 1 keV. Charged-

TABLE I. Age, gender, size, and voxel descriptions of the three models used in this study.

Model	Age	Gender	Weight (kg)	Height (cm)	Voxel size (mm)
Baby	8 weeks	Female	4.2	57	$0.85 \times 0.85 \times 4$
Child	7 years	Female	21.7	116	$1.54 \times 1.54 \times 8$
Irene	32 years	Female	51	163	$1.876 \times 1.876 \times 5$

particle equilibrium was assumed so that all the secondary electrons deposit their energy at the photon interaction sites.

II.B. MDCT source model

A 64-slice CT scanner system (Sensation 64, Siemens Medical Solutions, Forchheim, Germany) was modeled for all simulations using Monte Carlo-based methods. The widest available beam collimation of this scanner is 24×1.2 mm, which yields a nominal beam width of 28.8 mm. The corresponding actual beam width is 32.2 mm in air at the isocenter, which was evaluated from the measured radiation profile using optically stimulated luminescent dosimeters¹¹ (CT Dosimeter, Landauer, Inc., Glenwood, IL). The source file of MCNPX 2.6 was modified to model a Siemens Sensation 64 scanner such that helical scans with any clinical relevant kVp, mAs, pitch values, and collimation settings could be simulated.

The helical pathway of the CT source in x , y , z was explicitly defined for any scan given the pitch, table feed, tube start angle, nominal collimation, and scan start and stop location. The location of each emitting photon was then sampled from the specified pathway. The photon spectra for 120 kVp source and descriptions of any beam filtration, including the bowtie filter, as well as the geometry were provided by the manufacturer. As in previous work,¹³ this model was validated and benchmarked using comparisons based on standard dosimetry (CTDI) measurements and corresponding simulations, which agreed to within 5%.

II.C. Patient models

Several different patient models were evaluated and are described below. They are the Baby, Child, and Irene models from the GSF (now Helmholtz Zentrum München) family of computational, voxelized models^{21,22} as well as a limited sample of the pregnant patient models described in Ref. 16. In order to incorporate each phantom into the Monte Carlo code for particle transportation calculation, the element composition and mass density of each organ is required. These values were derived from the ICRU 44 organ composition tables.²³ For the GSF models, this included all radiosensitive organs, though only six are explicitly studied in this work; for the pregnant patient models, only the uterus, gestational sac, and fetus were explicitly studied; all other voxels were assigned to one of the six tissue categories (air, lung, fat, water, muscle, or bone) based on Hounsfield unit as described in Table I by DeMarco *et al.*²⁴

II.C.1. GSF phantoms

The GSF phantoms are a series of voxelized phantoms with segmented individual organs.²¹ Baby, Child, and Irene phantoms were selected from the GSF family to represent small patients with different body habitus. Some basic parameters of these three models are summarized in Table I. The selected organs used for evaluating dose variation were the testes, ovaries, breasts, thyroid, uterus, and eye lenses, whenever they are available in the phantom. These organs were selected because they are all relatively small in size so higher dose reduction is expected for these small organs. It should be noted that although Baby and Child are female phantoms, testes were added into the gonads by GSF in order to calculate average gonad doses. Also, despite the fact that Child is a female phantom, no breast voxels were identified.

II.C.2. Pregnant patient phantoms

In addition to GSF whole-body phantoms with various age and size, a series of previously developed pregnant female models¹⁶ with various gestational stages was used in this study to investigate the dose to fetus. The CT images of pregnant females were gathered from CT examinations performed under clinically indicated scans, such as trauma. The uterus, gestational sac, and fetus from these CT images were contoured by radiologists whenever they were visible. These three types of tissues were assigned as separate individual organs in the phantom and they were used to evaluate the dose variation caused by the tube start angle in CT scans. In total, there were 24 pregnant female models in this series of phantoms, from which 4 of them were selected for this study to represent different gestational stages, with emphasis on the early gestational stages because the size of the fetus is smaller. These four models are referred to as F26, F10, F7, and F31 in this study using the same nomenclature as in Angel 2008. The voxel size for each model is $2.4 \times 2.4 \times 5$ mm³. The fetal age, size of the uterus in longitudinal direction, and the studied regions for each model are summarized in Table II. The uterine length in the longitudinal direction for each model was estimated from the original voxel data.

II.D. Organ dose estimation

Under charged particle equilibrium condition, the absorbed dose to each voxel is equivalent to the collision kerma in this voxel, which was calculated by multiplying the mass energy-absorption coefficients with the energy fluence evaluated by MCNPX based on track-length estimation. Then

TABLE II. The size, gestational age, and related information for the studied pregnant female phantoms in this work (model refers to the patient model from Angel 2008).

Model	Gestational age (week)	Uterine length (cm)	Studied regions		
			Uterus	Gestational sac	Fetus
F26	5	7	√	√	—
F10	7	6	√	√	—
F7	12	7.5	√	√	√
F31	19	18.25	√	√	√

the mean dose to an organ was estimated by averaging the dose to each voxel across all the voxels belonging to the organ. All the organ dose results were converted from MCNPX raw output to absolute dose normalized to tube current (in mGy/mAs) using normalization factors calculated from scan measurements in air and corresponding simulations in air, described in a previous publication.¹⁴

II.E. Simulation experiments

For all the experiments in this study, simulated whole-body scans were performed for each phantom using 120 kVp and 300 mAs. Whole-body scan here refers to the full coverage of all the voxels in every phantom. For GSF phantoms, the scan limits extend from the top of the head to the bottom of the feet; for pregnant patient models, they extend from the lower thorax to the pubic symphysis.

To investigate the dose variation from different tube start angles, various tube start angles ranging from 0° to 340° with an interval of 20° were used for each set of simulations, which means 18 simulations were performed on each patient model using different tube start angles while keeping all the other scanning parameters constant. In the simulations, 0° means the tube is at the top of the gantry (12:00 position). To make the results more intuitive, the angle that is reported is not the start angle, but rather the angle of the x-ray source as it crosses the longitudinal center of the organ being investigated; which we will call the organ crossing tube angle (OCTA). For example, for the Baby GSF phantom, when using a pitch of 1.5 and a nominal beam collimation of 28.8 mm, a start angle of 0° (the angle of the source is at the 12:00 position at the start of the x-ray beam on), the angle of the x-ray source as it crosses over the center of the eye lenses is 107° (OCTA=107°). When the OCTA that results in the minimum dose for a particular organ is known, then the appropriate start angle can be calculated easily, as described below in Sec. II F.

To explore the correlation between the magnitude of organ dose reduction and the scanning parameters, pitch and collimation values were varied for each set of patient models. For the GSF models, three clinically relevant pitch values of 0.75, 1, and 1.5 were tested; for the pregnant patient models, only the pitch values of 0.75 and 1.5 were tested. For all patient models, two sets of collimation settings were simulated: One is the nominal beam width of 28.8 mm with the actual beam width of 32.2 mm; the other one is a case where the nominal beam width is 40 mm and the actual beam

width is 44.7 mm, calculated based on the ratio of the actual/nominal beam width for the 24×1.2 mm collimation. While 40 mm beam width is not currently available in the modeled scanner, many other manufacturers have scanners with the same or even larger beam widths.

For each patient model, organ of interest and set of scanning parameters (pitch and nominal collimation pair), the OCTA that results in the highest dose is used as the worst case reference. Based on this worst case reference value, the dose reduction from the worst case is calculated for each OCTA value. This value demonstrates how much savings is possible compared to the worst case. In the currently available MDCT scanners, there is no specific ability to control the start angle; therefore, the OCTA is similar to a random variable, with some patients receiving the highest dose to a specific organ and some the lowest dose to that same organ. For each experiment shown below, all results are presented as dose reduction compared to the worst case.

II.F. The calculations of OCTA

MATLAB subroutines were created to compute the OCTA at a given tube start angle, as well as to obtain the location, size, and centroid for each organ in each patient model. Given an organ's location and size information, the centroid along the longitudinal (*z*) axis is determined by averaging the longitudinal location of all the voxels assigned to the organ. When that centroid location is known as well as the scan start location (start of x-ray beam on, as described in Fig. 1) and the table feed per rotation (the product of pitch and nominal beam collimation), then the number of gantry rotations that will occur from the scan start location to the centroid of the organ was obtained by the following equation:

$$R = (\text{OLoc} - \text{SLoc}) / (\text{pitch} \times N), \quad (1)$$

where *R* is the number of rotations of the tube from the scan start location to the organ centroid, OLoc is the location of organ centroid along the longitudinal axis (*z* axis), SLoc is the longitudinal axis scan start location (where the x-ray beam first turns on), and *N* is the nominal beam collimation and pitch is conventionally defined. (Note that Pitch×*N*=table feed in mm/rotation.) After the calculation of *R* using Eq. (1), the OCTA is obtained by

$$\text{OCTA} = \text{tube start angle} + 360 \times (R - \text{floor}(R)), \quad (2)$$

where floor (*R*) is the integer part of *R*. It should be noted that the tube start angle that yields a particular value of

FIG. 3. The organ dose variation curves for Baby phantom from a simulated CT scan with various pitch and collimation settings: (a) Pitch 1.5 and 28.8 mm nominal collimation, (b) pitch 1 and 28.8 mm nominal collimation, (c) pitch 0.75 and 28.8 mm nominal collimation, (d) pitch 1.5 and 40 mm nominal collimation, (e) pitch 1 and 40 mm nominal collimation, and (f) pitch 0.75 and 40 mm nominal collimation. It should be noted that the same OCTA may correspond to different tube start angles for different organs.

OCTA for each organ is patient specific and organ specific. For example, an OCTA of 180° for the lens of the eye may have a different start angle from the one required to yield an OCTA of 180° for the thyroid. Similarly, OCTA of 180° for the lens of the eyes may correspond to different start angles for the Baby phantom and for the Child phantom.

III. RESULTS

III.A. GSF models

For each patient model, organ, pitch, and collimation combination, a set of 18 simulations was performed, each with different tube start angles. The organ dose reductions (compared to the worst case OCTA) are presented graphically by plotting the percent reduction in organ dose (on the y axis) as a function of the OCTA (on the x axis). Since three pitch values (0.75, 1, and 1.5) and two collimation settings (28.8 and 40 mm) were simulated, there are six such graphs

possible for each patient model. As an example, the dose reduction curves for all the studied organs in Baby phantom with all pitch values and nominal collimations are shown in Figs. 3(a)–3(f), with each figure demonstrating a different pitch/collimation combination. The organ dose reduction as a function of OCTA can be observed from these figures. For example, as shown in Fig. 3(d), the dose reduction compared to the worst case scenario for breasts for a scan with 1.5 pitch and 40 mm nominal beam width is almost 35%. For testes dose from the same scan, the dose reduction is 28%. It should be noted that the magnitude of dose variation is caused by solely changing the tube start angle while keeping all the other scan parameters constant.

These results indicate that for Baby, the largest dose reductions occur for eye lens, thyroid, breasts, and testes—all of which are at or near the surface of the patient and all of which have at least 20% dose reduction for pitch 1.5. These

TABLE III. The maximum dose reduction for all the studied organs for GSF phantoms under all simulated pitch values and collimation settings.

Organ	Collimation 28.8 mm			Collimation 40 mm		
	Pitch 1.5 (%)	Pitch 1.0 (%)	Pitch 0.75 (%)	Pitch 1.5 (%)	Pitch 1.0 (%)	Pitch 0.75 (%)
Baby breast	32	8	20	33	10	22
Baby ovaries	11	7	7	10	7	8
Baby testes	26	9	14	28	9	16
Baby thyroid	20	4	8	22	7	12
Baby uterus	8	5	5	9	6	7
Baby eye lens	38	3	15	41	6	21
Child ovaries	8	3	3	10	3	3
Child testes	33	6	11	36	9	16
Child thyroid	20	3	4	25	6	8
Child uterus	11	3	4	13	4	5
Child eye lens	49	7	20	51	8	23
Irene breasts	4	3	3	8	3	5
Irene ovaries	4	3	3	6	2	3
Irene thyroid	4	3	3	17	1	6
Irene uterus	4	2	2	8	2	2
Irene eye lens	52	11	17	54	13	22

results also show that the largest dose reductions occur for pitch 1.5 and pitch 0.75. Pitch 1 does not show large dose reductions, regardless of OCTA. This is not surprising because the magnitudes of the surface dose variations are much smaller when using pitch 1.0 than with pitch 1.5 or pitch 0.75, which is consistent with the results in phantoms as shown by Zhang *et al.*¹¹ These figures also show that the OCTA is different for different pitch values: It is approximately 180° for pitch 1.5 and approximately 0° for pitch 0.75. Finally, they also show slightly larger dose reduction for a wider beam collimation setting (40 mm compared to 28.8 mm).

Similar figures of organ dose reduction as a function of OCTA could be made for Child and Irene patient models. The behavior is similar and so they are not shown here; rather the maximum organ dose reductions for each patient model are summarized in Table III for all three GSF phantoms.

It can be shown from Table III that for the adult patient model (Irene), the dose reduction is not quite as large as the pediatric models except for the lens of the eyes. However, for the pediatric models (Child and Baby), the dose reduction to the breast, testes, thyroid, and eye lenses are all considerable (on the magnitude of 20% or more). Ovaries and uterus on pediatric patients have lower dose reduction (on the magnitude of 10%) because these two organs are located at depth (rather than at the surface) than the other four organs.

By comparing the results across different pitch values (different columns in Table III), it can be seen that the magnitude of dose reduction for pitch 1.5 is the largest, with the dose reduction up to 54% for Irene eye lens at 40 mm nominal beam width. For pitch 0.75, the magnitude of dose reduction is not as high as that for pitch 1.5, but it is still considerable, for example, 23% for Child eye lens at 40 mm nominal beam width. The dose reduction is not large for pitch 1, where it is below 10% for most of the organs in all GSF models.

By comparing the results across different collimations (28.8 and 40 mm in Table III), it can be seen that the magnitude of dose reduction for the 40 mm nominal beam width is slightly larger than that for the 28.8 mm nominal beam width for almost all of the cases. For example, for Child testes, the maximum dose reduction values at 28.8 mm nominal beam width for pitch 1.5, pitch 1.0, and pitch 0.75 are 33%, 6%, and 11%, respectively, while at 40 mm nominal beam width they are 36%, 9%, and 16%.

The comparison of the results across different organs on different patient models (different rows in Table III) is not as straightforward. In general, larger dose reduction occurs for smaller, peripheral organs in smaller patients. For example, the maximum dose reduction values of the eye lenses for all three phantoms are relatively large because the eye lens is a small surface organ. The maximum dose reduction for 40 mm collimation at pitch 1.5 (the fourth column in Table III) was plotted in a three dimensional space as a function of both organ length in the longitudinal dimension and the distance from the center of the organ to the isocenter. It is shown in Fig. 4, from which it can be observed that there is a general trend: As the organ size decreases and the distance to isocenter increases, the dose reduction will increase.

III.B. Pregnant female models

The organ dose variation curves for pregnant female models were plotted in the same way as for GSF phantoms. The

FIG. 4. The maximum dose reduction for individual organs from three GSF phantoms for a 40 mm collimation pitch 1.5 simulated CT scan as a function of both organ size in the longitudinal dimension and its distance to the isocenter in AP dimension.

organ dose variation can again be clearly observed. As an example, the percent dose reduction curves as a function of OCTA from the simulated scans with 40 mm nominal collimation, pitch 1.5 and pitch 0.75 for F7 phantom are shown in Fig. 5. The maximum dose reduction for pregnant model organs from the combinations of all the pitch and collimation settings is shown in Table IV.

This table shows that the magnitude of dose reduction to the fetus for pitch 1.5 can exceed 10% for the early gestational age models (≤ 12 weeks) for 28.8 mm collimation setting and increases to approximately 18% for 40 mm collimation setting. Results from pitch 0.75 yield smaller dose reductions with the earliest gestational age model, yielding only an 8% dose reduction for the 40 mm collimation setting. Results from the later gestational model (19 weeks) used in this study show very small dose reduction regardless of pitch, collimation, or start angle, which is to be expected for larger objects or organs.

This table also indicates that dose reductions are not as large as those observed in Baby and Child phantoms. This is understandable because the early gestational age fetus, while small, is located somewhat centrally in the body. This table does show some similar general trends between the dose reduction and the length in longitudinal dimension as in the GSF models in that a smaller size fetus has higher dose reductions. For example, at pitch 1.5, the maximum dose reduction is above 10% for F24, F10, and F7. However, for the

pregnant patient model with a later gestational age (19 weeks) the fetus is larger and the dose reduction is nearly negligible (F31). These results are also similar to that for GSF phantoms in terms of pitch and collimations: Pitch 1.5 and wider collimation yield larger organ dose reductions.

It should be noted that for the two phantoms with the earliest gestational age (F24 and F10), dose to fetus is not directly simulated in this study because the fetus tissue was too small to be identified by the radiologists from CT images; in the original study,¹⁶ the radiologist identified the gestational sac and we estimated dose to that tissue to serve as an indirect estimate of the dose to the fetus. This does not mean that there is no fetus dose; in fact, they may have a very high magnitude of dose variation from varying tube start angle because of their extremely small size. Furthermore, the development of fetus is very active in the initial few weeks so they are even more sensitive to radiation.²⁵

III.C. Computational time

The computational time for each MCNPX simulation (fixed pitch, collimation, and tube start angle) was about 5 min on a parallel computing cluster server with 64 AMD 2.0 GHz processors. Because for this study the requirement of the relative errors of the output results is very high, a total of 100 million photon histories were used for each simulation. The standard deviation of all the results from MCNPX was within 1.2%, with most of them below 0.5%. The scenarios where the error is higher than 0.5% happens when the size of the organ is very small, for example, eye lenses, because for such cases the number of photons that enters the organ can be small.

IV. DISCUSSION AND CONCLUSION

In this work, we have used Monte Carlo simulations to investigate the effect of the tube start angle on organ dose in several patient models. While current MDCT systems do not allow the user to control the start angle, this study investigated the feasibility of exploiting this sinusoidal behavior to reduce the dose to a specific organ by varying the tube start angle, especially for a small peripheral organ on small patients. Results were presented, showing the magnitude of organ dose reduction possible if the ability to choose the tube start angle were to be provided.

TABLE IV. The maximum organ dose reduction for several pregnant female phantoms under all the simulated pitch values and collimation settings.

Organ	Collimation 28.8 mm		Collimation 40 mm	
	Pitch 1.5 (%)	Pitch 0.75 (%)	Pitch 1.5 (%)	Pitch 0.75 (%)
F24 (5 weeks) gestational sac	12	5	17	8
F10 (7 weeks) gestational sac	12	5	18	6
F7 (12 weeks) fetus	11	2	18	5
F31 (19 weeks) fetus	1	1	3	3

FIG. 5. The organ dose variation curves for F7 phantom from a simulated CT scan with 40 mm nominal collimation and various pitch values: (a) Pitch 1.5 (b) pitch 0.75.

The results demonstrate a range of dose reduction for various organs in various phantoms undergoing a full body scan. This dose reduction can be large; for example, for Baby phantom the maximum reduction in the breast dose could be as high as 33%. For pediatric patients, the dose reduction to breast, testes, thyroid, and eye lenses is considerable (20% or more), while the dose reduction to ovaries and uterus is lower (10%). For adult patients, the dose reduction is only considerable for the eye lenses. For pregnant patients at early gestational stage (≤ 12 weeks), the dose reduction to fetus can be up to 17%–18%.

The organ dose variation resulted from varying the tube start angle depends on the scan parameters, especially pitch and nominal collimation. In a previous study,¹¹ it was shown that if the scanned object is a large (32 cm diameter) cylindrical homogeneous phantom, the dose distribution at the isocenter would be uniform no matter what the pitch and nominal collimation are. However, if the region of interest is at a peripheral or surface position (for example, breasts), instead of the isocenter, the longitudinal dose distribution

could have large variations. This is because at the peripheral positions, some areas are exposed to entrance, exit, and scattered radiation, while other areas may be exposed to either only exit and scattered radiation (e.g., pitch 1.5) or they may be exposed to an overlap region where they receive entrance radiation from two successive rotations (e.g., pitch 0.75).

The magnitude of organ dose reduction resulting from varying tube start angle is highly related to the location of the organ, size of the organ, and anatomy in the vicinity of the organ in question. If the organ is perfectly at the isocenter, the organ would have very minor dose variation from varying tube start angle because the dose distribution at isocenter is nearly uniform. If the organ is relatively large, such as lung or liver, it also will not have significant variation because its dimension in the longitudinal direction is much larger than the period of the dose distribution (which is basically table feed per rotation), and the organ dose integrated over several periods will not be able to reflect the peaks and the valleys. For example, the organ dose of F31 pregnant patient phantom does not have considerable variations de-

pending on the tube start angles because the organ (fetus) is relatively large. The anatomy in the vicinity of the organ in question is also an important factor. For example, the eye lens is surrounded by more bone than the breasts, so it has a higher magnitude of dose variation. In addition, although fetus is almost at the isocenter, it may be surrounded and protected from radiation by pelvis, except for the anterior direction. This will also result in some dose variation as a function of tube angle.

Since the table feed per rotation is the product of pitch and nominal collimation, these two parameters determine how “continuous” the surface (or peripheral position) is covered by the primary beam. For example, if the table feed per rotation is much larger than the actual beam width (as in pitch larger than 1.0), there may be considerable area on the surface that is not covered by the primary entrance beam and hence the longitudinal dose distribution will have higher variation. So theoretically, as the pitch and nominal collimation increase, the organ dose would have larger dose reduction. This was illustrated in Table III where the maximum dose reduction is higher for pitch 1.5 than that for pitch 1 for all cases, and it is slightly higher for 40 mm nominal beam width than that for 28.8 mm nominal beam width.

However, the dose reduction in pitch 1 is not higher than that for pitch 0.75. This is because as the pitch decreases to a certain value, the overlap of the adjacent radiation beams occurs, resulting in an overlap peak. As illustrated in Fig. 6, this scenario creates a new mechanism of organ dose variation, where inside the overlap region the organ receives higher dose and outside the overlap region the organ receives lower dose. Therefore, there is a “threshold” pitch for each organ in each phantom, such that the larger the difference between the pitch used in scan and the threshold pitch is, the higher the achievable dose reduction could be. It is expressed as

$$P_{\text{threshold}} = \left(1 - \frac{d}{\text{iso}}\right) \times \frac{A}{N}, \quad (3)$$

where d is the distance between organ centroid to the isocenter in the A-P dimension, A is the actual beam width, N is the nominal beam width, and iso is the distance between the x-ray tube anode to the isocenter. Typically, d (less than 10 cm for most of the organs) is much smaller than iso (usually 50–60 cm), so the threshold pitch can be approximated by A/N , which is about 1.1 in this study.

When the pitch used in the scan is higher than threshold pitch, cold spots are created along the longitudinal direction of the organ (as illustrated in Fig. 1). In comparison, when the pitch used in the scan is lower than threshold pitch, then overlap peaks result and “hot spots” are created as demonstrated in Fig. 6. If an organ has a relatively small extent in the longitudinal direction, its dose may be affected by these hot or cold spots, the location of which can be affected by the tube start angle. Thus two organ dose reduction strategies can be applied for these two different scenarios. For the first scenario (pitch $P > P_{\text{threshold}}$), the dose to an organ in question can be minimized if the tube start angle were to be

FIG. 6. Diagram illustrating the tube position, corresponding surface dose distribution along the z axis under two scenarios with different tube start angles, and a patient, at pitch 0.75. It was shown that the dose to the breasts of the patient can be affected by selecting different tube start angles.

selected in such a way that when the tube passes the organ, it is at posterior side of the patient (assuming the organ is on the anterior side of the patient); this corresponds to OCTA of 180° . For the second scenario (pitch $P < P_{\text{threshold}}$), the organ dose can be minimized by selecting a tube start angle such that when the tube passes the organ, it is at the anterior side of the patient, which corresponds to an OCTA of 0° . Of course, the effectiveness of both techniques also depends on the anatomy in the vicinity of the organ in question. From Fig. 3 it is seen that the optimal angles are slightly off by about 20° . One reason may be that the Baby model used in this study is in supine position slightly rolled to its left.

As the cone angle increases, the area of these cold or hot spots will also increase for helical scans, which causes larger variation in organ dose, depending on the tube start angle. Since very wide nominal beam width (>40 mm) MDCT scan systems are being introduced, this method could potentially serve as an important dose reduction technique.

Most of the investigated organs in this study are relatively small in size and easy to be anatomically located (e.g., lymph nodes are hard to be located). These organs happen to be located close to the anterior surface of the patient body. The

dose reduction strategies would have been different if these small organs were located at the back or at the side of the body. However, all the posterior and lateral organs have larger size in longitudinal direction comparing to the nominal collimation and hence the dose strategies being proposed in this study cannot be applied anyway.

The voxel sizes of all phantoms are considered small enough to accurately represent the selected organs, with the possible exception of eye lenses. For GSF Baby and Irene, there are 69 and 84 voxels identified as eye lenses, respectively. However, for GSF Child model, there are only 28 voxels identified as eye lenses. Given the voxel size for GSF Child ($1.54 \times 1.54 \times 8 \text{ mm}^3$), there are not a large number of voxels to represent the eye lenses, which may make it difficult to estimate the dose reduction accurately. However, because the beam width used here has a width of 28.8 mm (or 40 mm) and when pitch is great than one, there are large gaps between adjacent rotations; therefore, we would still expect significant dose reduction for the eye lenses.

There are several limitations in this study. First, the sample size of the patient models is relatively small. While the three GSF phantoms and the four pregnant female phantoms do represent a distribution of the patient size, they cannot represent the whole patient population. Since the magnitude of the dose reduction depends on the size of the organ, the dose reduction should have larger magnitude for smaller sized patients. But GSF Baby (8 weeks) is very small, so larger magnitude dose reductions would not be expected. For a larger size patient, the dose reduction may be lower than that observed in GSF model Irene, but perhaps not much lower, because larger patients do not necessarily have much larger organs, e.g., eye lenses. Second, this technique to reduce organ dose by selecting a specific tube start angle can only ensure that one organ gets the minimum dose while not controlling the dose to other organs. However, this does not impair this technique to be a useful organ dose reduction strategy because it does not put other organs at any higher risk than they are currently experiencing under uncontrolled (i.e., random start angle) helical scans. In addition, the simulations in this study considered constant tube current only. The effect from TCM is not simulated. This study did not specifically address any issues of image quality, though the impact would be expected to be insignificant because there is no change in acquisition parameters such as tube current, rotation time, beam collimation, or pitch. Also in order to apply the technique of reducing dose to certain organ by controlling the tube start angle, in practice, there may be some other practical considerations to take into account. For example, it requires the knowledge of the location of the organ prior to the CT scan (which could be available from the scan projection radiograph or planning view); the localization of the organ position should be very accurate; the organ must not move during the examination, etc.

In the currently available commercial CT scanners, the tube start angle is not controlled by the user and is either a completely random variable (any start angle is possible) or a limited random variable (only certain start angles are al-

lowed, but are still not controlled by the user). So the dose received by the organs investigated in this study using current MDCT scanners would be randomly distributed on the dose reduction curves shown. The purpose of this study was to investigate the magnitude of organ dose reduction that might be achieved if the capability to control the tube start angle were to be provided. This technique may be applied to a small organ on small patients, fetus on pregnant patients with early gestational stage, or eye lenses on adult patients. Since TCM techniques do not perform very well for pediatric patients (because they are more circular than elliptical, for example), this method could also serve as a complementary approach to TCM. The clinical realization of this methodology might not be trivial and would require implementation by scanner manufacturers; however, there are some current products which do at least take the tube start angle into consideration such as: (a) A dose reduction method which turns the x-ray beam off over the anterior portion of the patient (X-care, Siemens Medical Solutions, Forchheim, Germany) and (b) a method not used for radiation dose reduction but for synchronizing helical source paths between pre- and post-contrast scans (typically within the head) for the purposes of minimizing artifacts when subtracting the two scans (Sure Subtraction, Toshiba Medical Systems, Inc., Otawara-shi, Japan). These two products provide at least some indication that controlling the start angle may be feasible in some MDCT scanners in the future.

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