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Absorbed Radiation Dose in Radiosensitive Organs During Coronary CT Angiography Using 320-MDCT: Effect of Maximum Tube Voltage and Heart Rate Variations

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Abstract

OBJECTIVE—The purpose of this article is to estimate the absorbed radiation dose in radiosensitive organs during coronary MDCT angiography using 320-MDCT and to determine the effects of tube voltage variation and heart rate (HR) control on absorbed radiation dose.

MATERIALS AND METHODS—Semiconductor field effect transistor detectors were used to measure absorbed radiation doses for the thyroid, midbreast, breast, and midlung in an anthropomorphic phantom at 100, 120, and 135 kVp at two different HRs of 60 and 75 beats per minute (bpm) with a scan field of view of 320 mm, 400 mA, 320 × 0.5 mm detectors, and 160 mm collimator width (160 mm range). The paired Student's *t* test was used for data evaluation.

RESULTS—At 60 bpm, absorbed radiation doses for 100, 120, and 135 kVp were 13.41 ± 3.59, 21.7 ± 4.12, and 29.28 ± 5.17 mGy, respectively, for midbreast; 11.76 ± 0.58, 18.86 ± 1.06, and 24.82 ± 1.45 mGy, respectively, for breast; 12.19 ± 2.59, 19.09 ± 3.12, and 26.48 ± 5.0 mGy, respectively, for lung; and 0.37 ± 0.14, 0.69 ± 0.14, and 0.92 ± 0.2 mGy, respectively, for thyroid. Corresponding absorbed radiation doses for 75 bpm were 38.34 ± 2.02, 59.72 ± 3.13, and 77.8 ± 3.67 mGy for midbreast; 26.2 ± 1.74, 44 ± 1.11, and 52.84 ± 4.07 mGy for breast; 38.02 ± 1.58, 58.89 ± 1.68, and 78 ± 2.93 mGy for lung; and 0.79 ± 0.233, 1.04 ± 0.18, and 2.24 ± 0.52 mGy for thyroid. Absorbed radiation dose changes were significant for all organs for both tube voltage reductions as well as for HR control from 75 to 60 bpm at all tube voltage settings (*p* < 0.05). The absorbed radiation doses for the calcium score protocol were 11.2 ± 1.4 mGy for midbreast, 9.12 ± 0.48 mGy for breast, 10.36 ± 1.3 mGy for lung, and 0.4 ± 0.05 mGy for thyroid.

CONCLUSION—CT angiography with 320-MDCT scanners results in absorbed radiation doses in radiosensitive organs that compare favorably to those previously reported. Significant dose reductions can be achieved by tube voltage reductions and HR control.

Keywords

320-MDCT; absorbed radiation dose; cardiac imaging; cardiac MDCT; CT dose index; dose-length product; metal oxide semiconductor field effect transistor dosimetry

Radiation dose has been the single most significant concern in the context of widespread clinical applications of MDCT angiography. As a reference for radiation exposure, it should be noted that, in the general population, background exposure accounts for 82% and artificial exposure accounts for 18% of the total annual radiation exposure. Artificial exposure includes diagnostic x-ray studies, such as CT of the chest and abdomen, and the use of MDCT angiography [1]. Previous studies using 16- and 64-MDCT scanners often have not focused on absorbed radiation dose to radiosensitive organs but have used the dose-length product or weighted CT dose index ($CTDI_w$) and its derivatives, which are displayed by the equipment, for evaluation of radiation doses [2–4]. In addition to considerable doubts about the accuracy of these displays, these parameters are not suitable for evaluating absorbed radiation doses of specific organs that may be affected by primary beam or scatter radiation caused by the MDCT angiography examination [5–7].

Parameters such as the maximum tube voltage, retrospective image acquisition, heart rate (HR), and tube current time-product significantly affect the absorbed radiation dose associated with MDCT angiography [8, 9]. In addition, the effect of maximum tube voltage variation on radiation dose depends on the type of CT scanner because of individualized internal scanner filtrations and geometry. Also, results of maximum tube voltage variations have generally been evaluated for the CTDI or $CTDI_w$, but not as they pertain to specific organ doses [10]. Another parameter that influences absorbed radiation dose associated with MDCT angiography is HR. The objectives of this study were to measure the absorbed radiation dose in radiosensitive organs (i.e., chest and neck) during MDCT angiography using the 320-MDCT scanner with prospective gating and to evaluate the effect of changes in HR and maximum tube voltage on the absorbed radiation dose.

Materials and Methods

Study Design

Institutional review board approval was waived because no patients or animals were involved in this study.

All MDCT scans were performed with a 320-MDCT scanner (Aquilion ONE, Toshiba). The specifications and mode of operation of the scanner are described in detail in Appendix 1. All measurements were performed in a single session using an anthropomorphic phantom (RANDO, Alderson) and a high-sensitivity metal oxide semiconductor field effect transistor (MOSFET) radiation detector (model TN-1002 RD, Best Medical Canada, Ltd.) (Fig. 1). The anthropomorphic phantom is divided into 1-inch slices (2.54 cm) extending from the level of the cricoid cartilage to the lower pelvis, including slices 10 through 35. The starting point of the volume scan and coverage area was placed between slices 14 and 20, covering 160 mm in the z -axis. Four dosimeters were placed at slice 16, representing defined locations within the lung, which were constant throughout all measurements and included the dorsal, ventral, lateral, and left lung hilum. One dosimeter was placed at the left thyroid lobe, and one was placed at the lens of the eye. Two additional dosimeters were placed at slice 15, representing the breast surface, and in the midbreast. All dosimeters were secured in place (Fig. 2).

All scan protocols are based on the same scout view to ensure the same volume of coverage, thus standardizing measurement volumes among the three scan protocols. All data were collected during a single session using the same scan field of view (FOV) and scan range to ensure consistent data collection geometry. The phantom was then subjected to MDCT angiography with a scan FOV of 320 mm; 400 mA; 65% to R-wave default padding; 0.35-second rotation; 1.35-second total scanning time; 0.5-mm scan slice; 160-mm collimator width (160-mm range); half reconstruction; 175-milliseconds time resolution; and effective

140 mAs at 100, 120, and 135 kVp. Five scans were obtained for each maximum tube voltage value at a simulated HR of 60 beats per minute (bpm). Subsequently, the dose measurements were performed at 75 bpm with the same phantom setup but using 60% to R-wave default padding, 0.375-second rotation, 2.089-second total scanning time, 0.5-mm scan slice, 160-mm collimator width (160-mm range), segmented reconstruction, 99.65-millisecond time resolution, and effective 188 mAs. An ECG simulator, an integral part of the Toshiba equipment, was plugged into the CT scanner and was used for HR simulation and variation (Tables 1 and 2).

The mean of the five measurements for each detector at each HR and maximum tube voltage was converted into the organ equivalent dose by multiplying it with the appropriate tissue weighting factor for each organ (0.12 for lung and midbreast, 0.01 for breast [skin], and 0.04 for thyroid) [11, 12]. In addition, five scans used for calcium scoring (75% phase, 120 kVp, 400 mA, 0.35-second rotation time, 160-mm range, 140 effective mAs, and scan FOV of 320 mm) were performed with identical detector positioning. Because calcium scoring is an integral part of MDCT and MDCT angiography protocols at our institution at 300 mA, all values obtained at 400 mA were back-calculated to 300 mA. This type of calculation is routinely performed because the relationship between dose and milliamperage is linear in protocols that use prospective gating, such as the calcium score protocol. The dose for each organ was calculated on the basis of the five measurements obtained for each organ (Table 3 and Table 1 in Appendix 2).

Radiation Detectors

Before any dosimetry measurement is performed, MOSFET detectors require proper calibration matched to the primary beam quality of the CT scanner. Because the CT scanners in our department have beam qualities in the range of 6.5–7.5 mm aluminum half-value layer (HVL) at 120 kVp, detectors were calibrated with a conventional radiographic unit at 118 kVp with beam quality of 7.15 mm of aluminum HVL by adding 5.5 mm of aluminum to the faceplate of the collimator. The MOSFET detectors were subjected to radiation doses of 1, 3, 10, and 30 mGy. Individual calibration factors were obtained for all detectors by fitting these four data points with the least-squares fit. The software program used for the least-square fit was XLGENLINE (version 1.0, National Physical Laboratory). The conversion factors were then stored in the MOSFET software (mobileMOSFET software, version 2.0, revision 7.0, Thomson-Nielsen) for immediate readout after each protocol had been performed.

Direct calibration data entry using MOSFET's built-in calibration capability can also be used. However, to accommodate multiple tube potential calibrations, the least-square fit method was selected to verify that a given set of calibration factors obtained at 118 kVp with beam quality of 7.15 mm Al HVL may be used for different tube potentials with reasonable accuracy (i.e., within 5%).

Study Phantom and Detector Positioning

The anthropomorphic phantom consists of consecutively numbered individual transverse slices so that the phantom can be disassembled. In addition, the phantom contains longitudinal holes that traverse the entire phantom. These features allow MOSFET detector positioning for lung measurements within the phantom and intrapulmonary detector positioning for lung measurements (Fig. 2). The phantom head and trunk, but not the neck portion, were available for this study. An appropriately sized and shaped block of human tissue-equivalent material that is routinely used for CT dosimetry (standard polymethyl methacrylate, Plexiglas, Arkema France) was therefore placed between the phantom trunk and head and taped to the adjacent phantom parts as a neck surrogate.

A breast phantom (model 325 mL M, Mentor) was taped to the anterior chest wall. A detector was placed on the surface and within the breast phantom to simulate radiation dose that would be absorbed by mammary tissue.

To simulate coronary calcium scoring, the phantom was initially scanned using 120 kVp, 400 mA, rotation time of 0.35 second, scan FOV and display FOV of 320 mm, slice thickness of 3 mm, and scan volume equivalent to 160 mm coverage. Constant position of the dosimeters was maintained throughout all scans at the left lung hilum; dorsal, ventral, and lateral midlung; as well as the breast, thyroid, and eye. All MOSFET detector cables were secured with light tape to avoid detector movement during measurement acquisition.

Measurement Tabulations and Statistics

Measurements obtained with MOSFET detectors were entered simultaneously with image acquisition into an Excel database 2003 (Microsoft), and absorbed organ doses were tabulated in milligrays. Statistics were performed with SPSS for Windows (version 8, SPSS). Comparisons were later made for each organ dose, for HRs at 60 and 75 bpm with exposures at 100, 120, and 135 kVp, using the Student's paired *t* test. All scan parameters are included in Appendix 2.

Results

Mean absorbed organ doses were similar for all measurements within the primary beam (Table 2). The mean absorbed dose secondary to scatter radiation (thyroid absorbed dose) was substantially lower (e.g., 0.37 ± 0.14 mGy at 100 kVp, 0.69 ± 0.14 mGy at 120 kVp, and 0.92 ± 0.2 mGy at 135 kVp at 60 bpm) (Table 2 and Fig. 3B). For both HRs, mean absorption doses for all organs decreased significantly ($p < 0.05$) when maximum tube voltage was reduced from 135 to 120 kVp and from 120 to 100 kVp (Table 2 and Fig. 3C). Dose reduction was likewise significant for all organs when HRs were decreased from 75 to 60 bpm at all measured maximum tube voltage levels ($p < 0.05$). HR control from 75 to 60 bpm resulted in the greatest average absorbed radiation reduction across all organs (62%), followed by maximum tube voltage reduction from 135 to 100 kVp (55%) and from 120 to 100 kVp as well as 135–120 kVp (both 28%; Table 2). Averaged results obtained from calcium score protocols were as follows: midbreast, 11.2 ± 1.4 mGy; breast, 9.12 ± 0.48 mGy; lung, 10.36 ± 1.3 mGy; and thyroid, 0.4 ± 0.05 mGy. Measurements were obtained for the lens of the eye; however, the data were not recorded because measurements were so low that high SDs and inconsistencies would not allow a meaningful analysis of these particular data points (Table 3 and Fig. 3A).

Discussion

In this study, we used prospective gating with data acquisition in one rotation in which the tube is prospectively activated for only the phase window requested in the protocol. With this protocol, milliamperage is zero, except for the requested phase and a partial rotation on each side, to ensure enough data to reconstruct the first and last phase in the window (padding). We chose prospective gating for absorbed radiation dose measurement because it delivers the lowest dose for assessment of coronary artery disease. MOSFET detectors, which have been validated previously, were used for acquisition of absorbed radiation dose measurements and are characterized by increased ease of use compared with thermoluminescent dosimeters, the prior reference standard for point radiation dose measurements [13]. In addition to direct organ dose measurements, the volume CTDI (CTDI_{vol}) and the dose–length product have been used to assess radiation dose associated with CT examinations, both of which are usually displayed by the CT scanner. Even though CTDI_{vol} is widely used and well defined, it has substantial limitations because it does not

measure organ dose directly. In addition, it has been suggested that the $CTDI_{vol}$ may underestimate the absorbed radiation dose by 20–30% [6, 7, 11, 12]. Indeed, even proponents of the CTDI concede that its usefulness is limited to quality assurance and interequipment comparisons [14, 15]. Similarly, the dose–length product is a measure of the total exposure of the entire series of images. Although there is some controversy surrounding the accuracy of organ dose measurements derived from the dose–length product, there is consensus that it should not be used for individual patients or detailed risk assessment [11, 16]. This phantom study with direct dose measurements seems preferable to using surrogates of $CTDI_w$ and its derivatives.

In this study, absorbed radiation dose was measured for the most radiosensitive organs that are either within the primary radiation beam (lung and breast) or exposed to scattered radiation (thyroid gland). A small amount of osseous material (bone marrow) lies within the primary beam and is therefore subjected to some degree of stochastic radiation. Thus, absorbed radiation dose in the sternum is well approximated by the chest wall measurement. Other organs within the primary beam, such as neural, cardiac, and esophageal tissue, are relatively radioresistant. Additional dose measurements addressing these anatomic structures were therefore not conducted in this study.

Our data show significant average dose reductions caused by maximum tube voltage decreases from 135 to 100 kVp (55%) as well as from 120 to 100 kVp and from 135 to 120 kVp (both 28%). In addition to reducing dose, lower tube voltage generally has a positive effect on image quality because lower tube potential enhances subject contrast while contributing less Compton scatter [3]. HR control from 75 to 60 bpm reduces absorbed radiation dose by, on average, 62%. Although it will hardly be possible to decrease the maximum tube voltage to 100 kVp or the HR to 60 bpm in all cases, the aim should be for the lowest possible maximum tube voltage and HR individualized for each patient. The benefits of lowering HR to improve image quality and reduce radiation dose have previously been emphasized [17]. Specifically, lower HRs will prolong segments of the cardiac cycle that are relatively devoid of motion (e.g., end-diastolic relaxation and end-systolic contraction), which are therefore useful for image reconstruction [18]. However, even if adequate image quality can be achieved at higher HRs, HR reduction is advised for radiation reduction in each individual case. Our study was designed to compare the measured dose rate to the dose rates using dose–length product from the clinical scanner. The study was designed to compare radiation dose between the variations in heart rate and maximum tube voltage and not for image quality and interpretability for diagnosis.

The measured dose with a heart rate of 60 bpm and 120 kVp was 19.09 ± 3.12 mGy, and use of the conversion coefficient factor 0.12 resulted in 2.29 mSv, while the data set was collected in one heartbeat. With heart rates greater than 65 bpm, the data set is captured in two beats, thereby increasing the mean dose to 58.89 ± 1.68 mGy, or 7.07 mSv. Our data are comparable to those reported in a clinical study by Rybicki et al. [19] who used patients comparable to our phantom size (body mass index [BMI] < 30) with similar exposure factors, resulting in an average dose of 6.7 ± 1.4 mSv at 120 kVp in one heartbeat and 12.64 ± 3.4 mSv with two heartbeats. The apparent difference is most likely related to the method used to calculate the dose because Rybicki et al. used a 300-mm ion chamber and a scaling factor for dosimetry suggested by Mori et al. [20], which was then applied to the 320-MDCT scanner, or a slight increase in width of the phase window (padding). In a phantom study similar to ours, using MOSFET detectors with 75% phase window, Einstein et al. [21] compared the dose at 120 kVp to that at 100 kVp and reported a reduction in mean radiation dose from 2.19 to 1.12 mSv, which is similar to our results of a reduction from 2.29 mSv to 1.46 mSv at similar maximum tube voltage values. In a clinical study, Achenbach et al. [22] used a dual-source 64-MDCT system and reported a mean effective dose of 0.87 ± 0.08 mSv

using 100 kVp with a heart rate of less than 60 bpm. Lell et al. [23] also used a dual-source MDCT scanner and reported a mean effective dose of 0.88 ± 0.07 mSv using 100 kVp with a heart rate of less than 60 bpm. In our study using prospective gating, we also obtained corresponding organ equivalent doses values of 2.6 mSv for the midbreast, 0.19 mSv for the skin (breast surface), 2.29 mSv for the lungs, and 0.03 mSv for the thyroid at 120 kVp, which are much lower than data reported for retrospective studies [24].

Calcium scoring is usually routinely performed in addition to contrast-enhanced MDCT angiography with single rotations and prospective gating. However, our measurements suggest that unenhanced scanning for calcium scoring would increase the listed absorbed organ doses only modestly (Table 3 and Fig. 3A). Our measurements for the calcium score protocol at 120 kVp compare favorably to those of prior investigations using prospective gating [25, 26].

These comparisons reveal that MDCT angiography with the 320-MDCT scanner can be performed within one R-R interval with good heart rate control while simultaneously reducing prior levels of individual radiation burden associated with the examination reported in retrospective studies [24]. In addition, further reductions can be achieved by reducing coverage in the z-direction from 160 to 120 mm. Our data, along with those of others, also indicate that the occurrence of deterministic skin effects secondary to MDCT angiography with the dual-source and 320-MDCT scanner is fundamentally inconceivable, even in the setting of performance of multiple studies in a short time span in the same individual, because deterministic effects have a threshold of 2,000–20,000 mGy, depending on the severity [27]. It is believed that transient erythema, the earliest radiation effect, is unlikely to occur with radiation doses of less than 4,000–5,000 mGy [27].

Our study has some limitations. It is difficult to compare our data to those of many prior reports, which have often simply referred to the CTDI or dose-length product displayed by the equipment or have listed total effective dose back-calculations based on those numbers. To facilitate comparability to prior data and data understanding, our absorbed organ dose measurements were converted into effective organ doses. At the same time, it should be noted that the use of the organ equivalent dose has also been questioned given the source of its calculation, which includes the introduction of organ as well as radiation weighting factors [28]. Indeed, the tissue weighting coefficient for the breast has been changed twice since 1991, attesting to the arbitrary nature of organ equivalent dose calculations [1].

Another limitation of our study is the lack of radiation data correlation to image quality or diagnostic accuracy assessment. Also, weight, height, and proportions of the phantom that we used are meant to reflect those of the average person, but the BMI in the average patient with coronary artery disease can be expected to be higher than the BMI of the phantom that was used. Hence, scatter incurring from the examination as well as the milliamperage that is required to perform the study may actually be higher in these patients than in our study, resulting in a higher total absorbed dose, particularly at the breast surface.

Dose measurements at the approximate location of the eye lens were also obtained and were magnitudes lower than those for the thyroid gland. However, these data were not evaluated because of high SDs and data inconsistencies. This is likely because MOSFET detectors lack measurement reliability for such low dose levels. Finally, we used a constant current value of 400 mA for all measurements, which is routinely used for performance of coronary CT angiography with this scanner at our institution. We realize, however, that lower milliamperage levels may be permissible in some patients, whereas higher levels may be necessary in others for diagnostic imaging, thus resulting in different absorbed organ doses. For prospective gating used in this study, absorbed organ doses are linearly proportional to

the milliamperere levels. This allows easy calculations of different milliamperere levels and makes additional radiation dose measurements unnecessary. On the other hand, other scanning protocols (e.g., tube modulation) would have a more complex effect on absorbed radiation dose and necessitate additional measurements.

In conclusion, the performance of coronary CT angiography with a 320-MDCT scanner results in absorbed radiation dose in radiosensitive organs that compare favorably to those cited in previous reports. Significant dose reduction is achieved by lowering maximum tube voltage from 135 to 100 kVp (55%), from 135 to 120 kVp (28%), and from 120 to 100 kVp (28%) and by decreasing the HR from 75 to 60 bpm (62%).

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APPENDIX 1

Basic Operation of Toshiba Aquilion ONE

Aquilion ONE is a cone beam MDCT with 320 rows of 0.5-mm detector arrays capable of covering 160 mm of anatomy in the z-direction in one rotation without table movement. However, Aquilion ONE has two operating modes, the 64-row and the 320-row modes. Under the 64-row mode, Aquilion ONE functions just like another 64-MDCT scanner in which both axial scans and helical scans are available with a maximum beam width of 32 mm (64×0.5 mm).

The 320-row mode, at present, functions primarily as a cardiac scanner (volume scan mode). Under the volume scan mode, the unit acts as an axial scanner and can cover a range from 40 mm up to 160 mm in 20-mm increments per rotation. For cardiac scanning, the beam width coverage is usually 120, 140, or 160 mm, depending on the length of the heart. The tabletop translation movement is disabled via software control during image acquisition

under the volume scan mode, the data set is acquired in one 360° rotation, and the image is reconstructed from 180° using the remaining 180° for cone beam correction.

The tube current of a CT scan may be modulated to minimize the radiation absorbed by the patients. Typically, the automatic tube current modulation is based on the attenuation information gathered during the scout view, or the prescan projection radiographs. Hence, depending on the scout view information, the tube current of a CT scan may be modulated as a function of tube location in the rotating plane and the patient's location coded in the z -axis. Thus, the tube current may be modulated as a function of tube angular location in θ within a given rotation for a range of distance in the z -direction. Thus, milliamperage can be expressed as $\text{mA} = \text{mA}(\theta, z)$.

In the figure, Inset A depicts the R-R interval of the cardiac cycle in seconds (abscissa), and the height of the shaded areas represents the tube current (mA). For the volume scan as depicted in Inset A, the applied tube current is at a constant preset (mA value) throughout the complete R-R interval. This scheme will result in a higher dose to the patient compared with the other three modified tube current schemes. In Insets B–D, the dose is therefore similar to retrospective gating in typical helical scanning and results in a higher radiation dose. However, images can be reconstructed throughout the cardiac cycle for functional analysis.

In our study, the coronary CT angiography is ECG gated, and the tube current modulation is synchronized with the heart beat. In the case of the calcium scoring (Inset B in the figure), the tube current is turned “on” at 75% of the R-R interval for heart rates less than 65 beats per minute (bpm) and 40% for heart rates greater than 65 bpm. Assuming the same tube current is applied and keeping all other scan parameters the same, the radiation dose delivered will be less than one quarter of the scheme in Inset A. Inset C depicts prospective gating because the tube current is on for only a fraction of the R-R interval and the dose rate is much lower. For heart rates greater than 65 bpm, extra exposure is added before and after the usual amount needed for heart rates less than 65 bpm and is called “padding.” Padding is added because several reconstructions on either side of 75% may be needed for good image quality. In the case of coronary CT angiography, as depicted in Inset C, the tube current is turned on between 75% for heart rates less than 65 bpm and 40% to 90% for heart rates greater than 65 bpm. Tube current is modulated and applied throughout the R-R interval for the cardiac perfusion and functional studies, as shown in Inset D. The tube current is increased between 75% and 90% of the R-wave for heart rate less than 65 bpm and between 30% and 90% for heart rate greater than 65 bpm.

APPENDIX 2

Calibration of Metal Oxide Semiconductor Field Effect Transistor (MOSFET) Radiation Detectors

First, a conventional three-phase constant potential radiographic unit was selected for the purpose of MOSFET dosimeter calibration. The radiographic unit was evaluated for the tube potential accuracy at 100, 120, and 135 kVp (140 kVp for other CT scanners). The radiographic technique factors required to produce these four tube potentials was recorded with the radiation output under a fixed geometry for later use. An electrometer (Barracuda, RTI) was used with the multipurpose detector for the measurements of tube potential, radiation output (milligrays), and the radiation beam quality (in “millimeters of aluminum” of half-value layer [HVL]). The Barracuda multipurpose detector dosimetry system was calibrated by the manufacturer.

MOSFET detectors require proper calibration matched to the primary beam quality of the CT scanner, and because the CT scanners in our department have beam qualities in the range of 6.5–7.5 mm Al (HVL at 120 kVp, MOSFET detectors were calibrated with the aforementioned radiographic unit at 118 kVp (measured tube potential = 120 kVp) with a beam quality of 7.15-mm Al HVL. The beam quality of 7.15-mm Al HVL was obtained by adding 5.5 mm of aluminum to the faceplate of the collimator.

The MOSFET detectors were subjected to radiation doses of 1, 3, 10, and 30 mGy, as measured with the Barracuda multipurpose detector dosimetry system. Individual calibration factors were obtained for all detectors by fitting these four data points with the least-squares fit. The software program used for the least-squares fit was XLGENLINE software (version 1.0, National Physical Laboratory). The conversion factors were then stored in the MOSFET software (mobileMOSFET software version 2.0, revision 7.0, Thomson-Nielsen) for immediate readout of radiation output. Thus, the MOSFET dosimetry system was then calibrated for 120 kVp and 7.15-mm Al HVL.

To assess the accuracy of the MOSFET dosimetry system at other tube potentials, the MOSFET dosimeters were irradiated with the same radiographic unit at 100-, 135-, and 140-kVp tube potentials with the 5.5 mm of aluminum attached to the collimator faceplate. The radiation dose (output) was compared against the readings of the Barracuda. Although the calibration was a 1-point calibration at 120 kVp and 7.15-mm Al HVL, reasonable accuracy (< 5%) at other tube potentials was confirmed. Thus, no additional calibration at tube potentials other than 120 kVp was performed.

TABLE 1

Parameters of MOSFET Scans

	Scan Parameters						
	Tube Potential (kVp)	Tube Current (mA)	Rotation Time (s)	Total Scan Time (s)	Total Beam Width (mm)	Range (mm)	Reconstruction
Calcium score							
Volume ^a	120	300	0.35	0.35	0.5 × 320	160	Half
Volume ^b	120	300	0.35	0.35	0.5 × 320	160	Half
Prospective CTA							

	Scan Parameters						
	Tube Potential (kVp)	Tube Current (mA)	Rotation Time (s)	Total Scan Time (s)	Total Beam Width (mm)	Range (mm)	Reconstruction
Scan and view ^c	100	50	0.5	0.5	0.5 × 4	2.0	Half
SureStart ^d	100	50	0.35	40.25	0.5 × 4	NA	Half
Volume ^e	120	400 (P)	0.35	1.35	0.5 × 320	160	Half
Scan and view	100	50	0.5	0.5	0.5 × 4	2.0	Half
SureStart	100	50	0.35	40.25	0.5 × 4	NA	Half
Volume ^f	120	500 (P)	0.375	2.089	0.5 × 320	160	Segmented
Cardiac CTA and CFA							
Scan and view ^c	100	50	0.5	0.5	0.5 × 4	2.0	Half
SureStart	100	50	0.35	40.25	0.5 × 4	NA	Half
Volume ^g	120	400 (M)	0.35	1.35	0.5 × 320	160	Half
Scan and view	100	50	0.5	0.5	0.5 × 4	2.0	Half
SureStart	100	50	0.35	40.25	0.5 × 4	NA	Half
Volume ^h	120	500 (M)	0.375	2.089	0.5 × 320	160	Segmented

Note—Field of view for all scans (modulated) is 320 mm. SureStart manufactured by Toshiba. CFA = computed fluoroscopic angiography, CTA = CT angiography, M = modulated, NA = not applicable, P = padded.

^a At 60 bpm, default (mA-control) is 75% of R-wave.

^b At 75 bpm, default (mA-control) is 40% of R-wave.

^c One axial scan and view image.

^d CT fluoroscopy: continuous scan up to a maximum of 40.25 seconds.

^e At 60 bpm, default (mA-control) is 65–90% of R-wave.

^f At 75 bpm, default (mA-control) is 30–90% of R-wave.

^g At 60 bpm, default (mA-control) is 75–90% of R-wave.

^h At 75 bpm, default (mA-control) is 30–90% of R-wave.

Fig. 1. Metal oxide semiconductor field effect transistor (MOSFET) detector system (model TN-1002 RD, Best Medical Canada, Ltd.) shows dosimeter (*black arrow*) containing silicone chip and FLEX (Kapton) insulated wiring that conducts current as surrogate for dose (*white arrow*), both of which were placed at previously defined locations on phantom surface, within anthropomorphic phantom (RANDO, Alderson) and breast phantoms and MOSFET connector (*gray arrow*).

Fig. 2. Images of phantom with metal oxide semiconductor field effect transistor (MOSFET) detectors in position

A, CT anteroposterior scout image. Numbers on right correspond to slice location of detectors embedded within or on surface of phantom. Arrows at left upper thorax indicate location within and on surface of breast. Arrowheads indicate detector location within dorsal, ventral, lateral, and left lung hilum at slices 17 and 18.

B, Coned axial view of phantom shows MOSFET detectors and their location from pleural surface dorsal (9.4 mm), ventral (35.9 mm), lateral (19.1 mm), and left lung hilum (101, 79, and 82 mm, respectively, from ventral, dorsal, and lateral pleural surfaces).

Fig. 3. Graphs show measured organ radiation doses

A, Graph shows measured absorbed organ doses for coronary CT angiography (CTA) and calcium scoring at 120 kVp.

B, Graph shows measured organ equivalent doses for coronary CTA at 60 beats per minute (bpm).

C, Graph shows measured organ equivalent doses for coronary CTA at 75 bpm.

TABLE 1

Scanning Parameters

Scan Protocol	R-Wave (%)	Rotation (s)	Reconstruction	mAs	Time Resolution (ms)	Scan Time (s)
Coronary CT angiography at 60 bpm	65	0.35	Half	140	175	1.35
Coronary CT angiography at 75 bpm	30	0.375	Segment	188	99.65	2.09
Calcium scoring	75	0.35	Half	140	175	0.35

Note—Other scanning parameters are as follows: slice thickness, 0.5 mm; current, 400 mA; scan field of view, 320 mm; and collimator, 160 mm (apply to all protocols).
bpm = beats per minute.

TABLE 2

Measured Absorbed Organ Equivalent Doses at 100, 120, and 135 kVp Acquired at Simulated Heart Rates of 60 and 75 Beats per Minute (bpm)

Heart Rate, Body Region	Tube Voltage (kVp)		
	100	120	135
60 bpm			
Midbreast	13.14 ± 3.59	21.7 ± 4.21	29.28 ± 5.17
Breast	11.76 ± 0.58	18.86 ± 1.06	24.82 ± 1.45
Lung	12.19 ± 2.59	19.09 ± 3.12	26.48 ± 5.0
Thyroid	0.37 ± 0.14	0.69 ± 0.14	0.92 ± 0.2
75 bpm			
Midbreast	38.34 ± 2.02	59.72 ± 3.13	77.8 ± 3.67
Breast	26.22 ± 1.47	44.00 ± 1.11	52.84 ± 4.07
Lung	38.02 ± 1.58	58.89 ± 1.68	78.00 ± 2.93
Thyroid	0.79 ± 0.23	1.04 ± 0.18	2.24 ± 0.52

Note—Data are mean ± SD absorbed dose (mGy). All scans were obtained with 160-mm collimation.

TABLE 3

Measured Absorbed Organ Equivalent Doses From Calcium Scoring at 120 kVp

Body Region	Absorbed Dose (mGy)
Midbreast	11.2 ± 1.4
Breast	9.12 ± 0.48
Lung	10.36 ± 1.30
Thyroid	0.4 ± 0.05