Practical dosimetry methods for the determination of effective skin and breast dose for a modern CT system, incorporating partial irradiation and prospective cardiac gating

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Objective: For CT coronary angiography (CTCA), a generic chest conversion factor returns a significant underestimate of effective dose. The aim of this manuscript is to communicate new dosimetry methods to calculate weighted CT dose index (CTDI_w), effective dose, entrance surface dose (ESD) and organ dose to the breast for prospectively gated CTCA.

Methods: CTDI_w in 32 cm diameter Perspex phantom was measured using an adapted technique, accounting for the segmented scan characteristic. Gafchromic XRCT film (International Speciality Products, New Jersey, NJ) was used to measure the distribution and magnitude of ESD. Breast dose was measured using high sensitivity metal oxide semiconductor field-effect transistors and compared to the computer based imaging performance assessment of CT scanners (ImPACT) dosimetry calculations.

Results: For a typical cardiac scan the mean ESD remained broadly constant (7–9 mGy) when averaged over the circumference of the Perspex phantom. Typical absorbed dose to the breast with prospectively gated protocols was within the range 2–15 mGy. The subsequent lifetime attributable risk (LAR) of cancer incidence to the breast was found at 0.01–0.06 for a 20-year-old female. This compares favourably to 100 mGy (LAR ~0.43) for a retrospectively gated CTCA.

Conclusions: Care must be taken when considering radiation dosimetry associated with prospectively gated scanning for CTCA and a method has been conveyed to account for this. Breast doses for prospectively gated CTCA are an order of magnitude lower than retrospectively gated scans. Optimisation of cardiac protocols is expected to show further dose reduction.

A number of publications have compared radiation doses between invasive angiography (IA) and CT coronary angiography (CTCA) [1–4]. For comparison between two differing radiographic modalities we must first convert the relevant dose descriptors to effective dose (mSv). It is not our intention to discuss the limitations of effective dose as these are published elsewhere, although the reader should be aware that error has been estimated as high as 40% when performing such calculations [5]. Nonetheless, to reduce error and promote fair comparison, it is important to be as accurate as possible when performing such calculations. Current scientific thinking is that effective dose should not be used as a definitive risk descriptor for small volume irradiations (such as CTCA) for individuals. Instead, specific organ doses should be estimated alongside age-categorised organ risk of cancer [6, 7]. The estimation of effective dose remains useful when comparing two separate imaging

modalities and the use of conversion factors is relatively simple. However, there have been significant developments in terms of both technique and equipment as well as the revision of tissue weighting factors. Therefore, there is a need to re-evaluate previously published conversion factors for estimation of effective dose for emerging radiological practices on new imaging systems. The dosimetry methods for CTCA detailed within this publication were utilised during the comparison of effective dose between prospectively gated CTCA and IA [8]. The methods described for CTCA dosimetry were also used to evaluate the use of generic conversion coefficients in the estimation of effective dose from CTCA [9]. Within the Plymouth Hospitals NHS Trust, 90% of the CTCA examinations are performed utilising prospectively gated CTCA and so the dosimetry methods have been adapted for this set of clinical protocols.

A number of authors [4, 10, 11] have estimated effective dose for CTCA. Some have compared image quality and radiation dose between the two imaging techniques [10] while others have looked at advances in the field of CT [3, 4, 11]. Very few authors have calculated effective dose for all patients in their study group [1]. Most

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Received 11 June 2010
Revised 8 October 2010
Accepted 29 November 2010

DOI: 10.1259/bjr/22285164

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Figure 1. Radiation profile from a single 40 mm axial slice for GE Lightspeed VCT 64-slice scanner (General Electric Healthcare, Milwaukee, WI) [Gafchromic® XR QA2 film (International Speciality Products, New Jersey, NJ) wrapped around exit port of gantry].

authors utilise previously established conversion factors applied to scanner-recorded dose length product (DLP). The chosen conversion coefficient is normally within the range of $0.014\text{--}0.019\text{ mSv (mGy} \times \text{cm)}^{-1}$ [2–4, 10, 12]. These conversion factors are based on studies of the whole chest using International Commission on Radiological Protection (ICRP) publication 60 tissue weighting factors [13]. Since the first use of such conversion factors in 1991, CT technology has changed significantly. Modern scanners incorporate a range of developments to enhance cardiac imaging and reduce radiation dose—for example, cardiac specific bow tie filters, dual energy imaging, prospective partial rotation gating, milliamp modulation and iterative reconstruction. Dosimetry methods are presented in this study to show the estimation of effective dose, entrance surface dose (ESD) and organ dose to the breast.

Advances in technology and scanning protocols have complicated the assessment of effective dose and dosimetry calculations for modern cardiac CT. We aim to evaluate and update dosimetry methods for prospectively gated CTCA with consideration of non-uniform irradiation, axial pitch, weighted CT dose index (CTDI_w) measurement technique modification, mean ESD, cardiac specific volumes and bow tie filters. Measurement of breast dose using metal oxide semiconductor field-effect transistor (MOSFET) detectors is compared with calculated breast dose using the Monte Carlo technique incorporated in the imaging performance assessment of CT scanners (ImPACT [15]) dosimetry calculator. Typical breast dose from clinical protocols will be used to estimate the lifetime attributable risk (LAR) of developing a solid cancer.

Methods and results

CT dosimetry measurements

CTCA effective doses were calculated using the ImPACT dosimetry calculator (incorporating ICRP 103

tissue weighting factors) version 1.0 [6]. All patients were scanned using either a GE Lightspeed VCT or GE HD750 (General Electric Healthcare, Milwaukee, WI) 64-slice cardiac CT scanner using a prospectively gated imaging protocol (SnapShot Pulse™, General Electric Healthcare). CT dose index measured in air (CTDI_{air}) was recorded at the isocentre of the radiation field. These values were entered into the ImPACT calculator to enable scanner-specific calculations to be performed. Measurements were performed by simulating the clinical techniques whereby the scanner was placed into a test mode and an electronically generated ECG signal could simulate a regular beating heart and enable cardiac modes at any heart rate. For all our measurements, the ECG input was set to 60 bpm, representing a perfectly stable heart at this cardiac frequency.

Non-uniform irradiation

Radiochromatic film (Gafchromic® XRQA2, International Speciality Products, New Jersey, NJ) was used to determine the proportion of beam on/off time during one 360° rotation. The film was wrapped around the exit port of the scanner gantry and scanned using the prospectively gated cardiac protocol. The film was later read within a CanoScan LiDE 200 (Canon, Ohta-ku, Tokyo) flat bed scanner and the dose profile plotted as a function of tube angle. The resulting profile may be seen in Figure 1. In this case no attempt was made to calibrate the film for radiation dose as we were only interested in the distribution and not concerned with absolute dosimetry. The film is relatively noisy, partly because the film has a printed calliper scale which affects the scanned profile.

Using this information and the time taken for the X-ray tube to complete one rotation (350 ms), we estimated the total “beam on” time for a complete rotation. Beam on/off proportion was found at 65% and 35% of the total gantry rotation, respectively. The exposure time was calculated as 227 ms ($0.65 \times 350\text{ ms}$) or 234° of a complete rotation. General Electric Healthcare state in their

Figure 2. Radiation profile from a typical 4-slab prospectively gated CT coronary angiography protocol (shown for GE Lightspeed VCT 64-slice scanner; General Electric Healthcare, Milwaukee, WI). Axial overlap apparent at joins.

literature that the X-ray beam is on for 180° (50% of a rotation). The deviation to our measured value of 65% may be accounted for when considering the radiation fan angle emerging from the X-ray tube. While the X-ray tube is only exposed for half a rotation, the emerging fan beam incident on the surface of a cylindrical phantom is 180° plus fan angle.

Axial pitch consideration

From a previous audit it had been established that 88% of CTCA requires 4 axial “slabs” of 40 mm collimated width in order to cover the heart (typical scan length of 140 mm). Radiochromatic film was again used to verify the X-ray dose distribution incident at the isocentre for a typical scan length (Figure 2). A 100 mm pencil ionisation chamber (Accupro, RadCal, Monrovia, CA) was used to calibrate the film. Results show clear overlap between each axial slice, the requirement of which is well described [14].

Figures 1 and 2 are important when considering modelling scanning technique using Monte Carlo phantoms (such as the ImPACT dosimetry calculator). Figure 2 indicates that there is a significant level of axial overlap at the isocentre [14] that will contribute to the overall estimate of effective dose. By analysing the area under Figure 2 it is estimated that the overlapping axial fields account for approximately 15% of additional dose. This additional dose was taken into account by using an overlapping “axial pitch” of 0.85 in the ImPACT dosimetry calculator.

Modification of $CTDI_w$ measurement technique

Currently all CT Monte Carlo simulators assume a complete rotation per axial slice, essentially assuming a known uniform distribution through the body/phantom described by the $CTDI_w$. For non-uniform irradiation, individual organ doses will change as a function of the start and end points of the X-ray tube. For the GE Lightspeed VCT and GE HD750 CT scanners evaluated, the start location of the X-ray tube rotational path is controlled by the scanner and is gated from the ECG to coincide with the next period of diastole and relative cardiac stasis. In essence, the irradiation path over the circumference of the

body is random. It has been assumed the irradiation path is random based on start/stop locations linked to the cardiac cycle and there are no other factors (for example, an in-built scanner delay). The dosimetric consequence of this is explained by considering the standard measurement of $CTDI_w$ in a 32 cm diameter phantom. Current methods for measurement of $CTDI_w$ involve the measurement of CTDI in a 32 cm diameter or a 16 cm diameter Perspex phantom (body/head) measured with a 100 mm pencil ionisation chamber at the centre and peripheral locations of the phantom. Typically this could be achieved by taking the average of two to three measurements at each scan location. When considering partial irradiation rotation techniques, the standard measurement technique for $CTDI_w$ must be adapted. Either we must now use numerous calibrated ion chambers and collect all peripheral measurements coincidentally or we must take the average of numerous peripheral measurements. This is owing to the randomised start and stop positions of the X-ray field incident on the patient’s or phantom’s surface. The scanner’s calculation of DLP was verified by first measuring $CTDI_w$ by averaging numerous CTDI measurements at central and peripheral locations in the 32 cm diameter Perspex phantom (Table 1). This is a modified technique from that described by the ImPACT Group [15]. The scanner displayed CTDI and DLP values were found to agree with our measured values within $\pm 5\%$.

CTCA and invasive angiography dose audit

The ImPACT dosimetry calculator (version 1.0 28/08/2009) [15] uses scanner matched data to allow the selection of appropriate Monte Carlo data sets to calculate effective dose. Users may adopt previous CTDI dose descriptors (namely $CTDI_{air}$ and $CTDI_w$) on identical scanners, manufacturers and models, or enter their own site/scanner specific measurements. When considering cardiac doses there is currently no existing library of $CTDI_w$ measurements suited to the use of the smaller General Electric bow tie filters, as used on the latest CT systems, which can be applied to a 32 cm diameter (body) dosimetry phantom. For this reason, $CTDI_{air}$ and $CTDI_w$ measurements were made locally (using the non-uniform irradiation dosimetry technique described

Table 1. Dose variation with 100 mm pencil chamber as function of position in 32 cm diameter Perspex phantom for a prospective CT coronary angiography protocol. The table shows high standard deviation (SD) in dose for peripheral locations compared with centre measurement. Results are shown for following scan protocol: 120 kV, 550 mA, cardiac medium bow tie filter, 64 × 0.625 mm detector configuration

Measurement #	Location of pencil ionisation chamber in Perspex Phantom				
	Centre	North	South	East	West
1	2.651	1.039	2.296	7.415	7.530
2	2.603	1.003	6.374	7.669	8.010
3	2.466	8.236	6.067	2.606	2.904
4	2.661	7.297	5.965	2.519	2.777
5	2.601	0.806	0.552	3.668	3.759
6	2.468	0.792	6.104	7.711	8.010
7	2.657	7.958	6.120	7.035	2.514
8	2.597	7.939	6.370	0.631	4.000
9	2.473	2.334	2.059	7.456	8.010
10	2.455	4.419	0.569	0.698	7.755
11	2.678	8.290	6.153	7.533	7.982
12	2.589	8.294	6.372	7.634	1.766
Mean dose (mGy)	2.575	4.867	4.583	5.215	5.418
SD (mGy)	0.086	3.425	2.427	2.932	2.637

above) on both the Lightspeed VCT and HD750 CT scanners matched to each protocol (namely kilovoltage and bow tie filter). This enabled the addition of our individual measured scanner and protocol specific values to the ImPACT dosimetry calculator. In doing this we were able to generate a set of “ImPACT factors” [16]. The ImPACT factors were then used to select a scanner-matched Monte Carlo data set, representing an appropriate model for the distribution of dose from our selected protocol.

An audit of recorded DLP was performed for both the GE Lightspeed VCT and GE HD750 CT scanners for prospectively gated CTCA at this site. Patient height and weight were recorded to allow dose audit to be restricted in line with principles within the National Protocol for Patient Dose Measurements in Diagnostic Radiology [17]. A body mass index (BMI) range of 20–35 was used in order to compare doses from a recent IA audit, thus allowing comparison between modalities (fluoroscopy *vs* CT). Our establishment adopts a fixed milliamp (mA) technique for a limited number of scan protocols according to patient BMI (Table 2).

For each of the 123 patients (84 filtered back projection, 39 iterative reconstruction) within the audit the effective dose was calculated using the scanner and technique specific CTDI doses using the ImPACT dosimetry calculator. Effective doses using ICRP 103 tissue weighting factors (mean, median and interquartile ranges) are shown in Figure 3 which is reproduced from Gosling et al’s research [8]. An incidental finding

Table 2. Prospectively gated cardiac protocol at this institution (shown for GE Lightspeed VCT 64-slice CT scanner, General Electric Healthcare, Milwaukee, WI)

BMI	kV	mA	CT Filter
<22.5	100	450	Medium
22.5–24.9	100	500	Medium
25–27.4	120	550	Medium
27.5–29.9	120	600	Large
30–34.9	120	650	Large
>35	140	650	Large

BMI, body mass index.

from this audit (using the dosimetry techniques within this publication) was a significant deviation from previously published DLP to effective dose conversion factors $[0.028 \text{ mSv} (\text{mGy} \times \text{cm})^{-1}]$ compared with $0.014\text{--}0.019 \text{ mSv} (\text{mGy} \times \text{cm})^{-1}$ [12, 18]. This agrees well with recent findings by Christner et al [19] and suggests a need to revise effective dose conversion factors for the chest to account for the change in tissue weighting factor for the breast [6].

Segmented scan considerations when calculating effective and organ doses in CTCA

Without a significant evolution of current CT Monte Carlo calculators for the estimation of effective dose in partial rotation CT we must assume that the radiation field is distributed uniformly around the body. This is clearly not the case and, although the effect may be diluted with averaging in cases where more than one axial partial rotation is performed, occasionally the dose may accumulate where trigger locations are synchronised. While this will have some effect on the calculated effective dose using existing Monte Carlo methods, the effect should be reduced when calculating the equivalent dose to all organs. This is not negligible when considering individual organ dose and will be further discussed with reference to the breast. The rare case, for example, where all four segments of the radiation field fall on the breast surface, will lead to a significant increase in both organ (breast) and effective dose, as CTCA encompasses a large proportion of breast tissue. This is further elevated by the recent increase in tissue weighting factors [6]. When estimating cancer risk from exposure scenarios in which different parts of the body receive substantially different doses, estimates of age- and sex-specific risks for individual tissues/organ are required [6, 7]. It is clear that the small scanning volumes encompassing the relatively radiosensitive breast tissue combined with high CT tube current indicate that the organ-specific radiation risk of cancer to the breast for CTCA should be calculated. From this discussion, clear advantage is gained if all radiation

Figure 3. Results from audit (invasive angiography vs CT coronary angiography). Reproduced from Gosling et al [8]. Methods from this communication were utilised within this audit. FBP, filtered back projection utilising a GE Lightspeed VCT 64-slice CT scanner, General Electric Healthcare, Milwaukee, WI; IR, iterative reconstruction with a GE HD750 64-slice CT scanner, General Electrical Healthcare.

arcs were to fall on the opposite side of the patient to the breast. This is one benefit of IA, where fluoroscopic examinations are undertaken with the X-ray tube under the table with the patient supine (thus significantly reducing the organ dose to the breast).

Entrance surface dose

To characterise our CT system for prospectively gated CTCA and to promote local understanding and optimisation of typical breast doses, Gafchromic® film XRQA2 was wrapped around the circumference of a 32 cm diameter Perspex phantom. The phantom was positioned at the isocentre of the tube rotation and exposed to a medium-size patient protocol for CTCA (120 kV, cardiac small filter, 550 mA, 350 ms tube rotation time). This film (post-irradiation) is shown in Figure 4. The film had been previously calibrated (Figure 5) for dose by exposing small pieces of film (from the same batch) to known doses at the surface of a 20 × 20 × 20 cm Perspex phantom over a suitable dose range (1.5–100 mGy). Doses were measured with the chamber on the phantom surface and aligned with the scanner's axis of rotation using both a 6 cm³ and a 3 cm³ CT ionisation chamber (Accupro, Radcal). The film used for calibration purposes was irradiated using the cardiac small bow tie filter and 120 kV over a suitable range of milliamperage values (10–700 mA) to achieve the desired dose range.

Following irradiation, the films were scanned using a flat bed scanner (CanoScan LiDE 200). An image manipulation tool (Image J version 1.41; Wayne Rasband, National Institute of Health, Bethesda, MD) was then

used to measure regions of interest over the calibration strips to enable a plot of pixel value *vs* chamber measured ESD. The fitted curve was then applied to the image from the film that was wrapped around the phantom to allow the ESD to be plotted along the length of the exposed film (Figure 6). For a regular cardiac cycle of 60 bpm, the mean ESD for a 4 segment (cardiac small filter, 550 mA, 120 kV, 140 mm) scan remained broadly constant (~7–9 mGy). We do not, however, know how the scanner reacts to an unstable heart rate where the ECG gating alters the natural timing of irradiation.

To evaluate the potential ESD range in dose from a single axial rotation, strips of radiochromatic film were exposed for all three cardiac filters (cardiac small, medium and large). The results are expressed graphically within Figure 7.

There would be clear dose benefits in only triggering the CT radiation arc under the patient. However, this may present technical challenges owing to the rapid rotation of the X-ray tubes (350 ms or quicker), the need to trigger the exposure during diastole and the need to acquire the complete examination in a single breath-hold. If this could be realised, it would lead to a significant dose saving to the patient.

Organ dose to the breast

Previous authors have not considered the effect of partial rotation on organ dose to the breast. High sensitivity MOSFET detectors (MOSFET 20, BEST Medical, Ontario, Canada) were first calibrated to relate

Figure 4. Gafchromic® XR QA2 film (International Speciality Products, New Jersey, NJ) used to measure entrance surface dose distribution for a typical prospective CTCA protocol. Film shown (a) pre- and (b) post-irradiation wrapped around circumference of 32 cm diameter Perspex phantom. (c) The post-irradiated film.

voltage readout to absorbed dose in a 32 cm diameter Perspex phantom using the 3 cm³ pencil ionisation chamber (Accupro, RadCal) and 40 mm axial slice for a complete rotation. Calibration factors for each of the 10 MOSFET detectors were calculated for kilovoltage/bow tie filter combinations typically used for prospective protocols. The ionisation chamber was positioned at the top (north) position, 1 cm from the phantom surface,

perpendicular to the axis of rotation (to encompass the whole chamber) and centred to the radiation field. The MOSFET detectors were positioned around the midline of the 10 cm pencil chamber for the calibration process. This was achieved by removing a 2 mm thick cylindrical sheath that normally surrounds the pencil chamber within the Perspex phantom during dosimetry measurements.

Figure 5. Calibration of Gafchromic film for entrance surface dose (ESD). Strips of film (identical batch) were exposed to increasing ESD measured with a 100 mm pencil ion chamber and a 6 cm³ ion chamber. Red channel pixel value (region of interest) was plotted as a function of ESD (chamber average). Power fit trend line used to estimate ESD for prospective protocols (CT coronary angiography).

Figure 6. Mean entrance surface dose (ESD) for cardiac small filter (12 V, 550 mA, SnapShot Pulse™ (General Electrical Healthcare, Milwaukee, WI), representing typical patient protocol). Film join is noticeable at two locations. Region of interest placed over complete length and width of film in Figure 4.

Five MOSFET detectors were then positioned in the middle of each breast within an adult anthropomorphic phantom (Atom® phantom, CIRS, Norfolk, VA) in a similar way to previous methods [20]. The position of the detectors is shown in Figure 8. The phantom was exposed to various prospective CTCA protocols (SnapShot Pulse™) on a GE HD750 CT scanner (500 mA, 350 s RT, 113.5 mAs). Owing to the varied nature of segmented irradiation, at least 3 measurements were made for each protocol (minimum of 15 measurements per breast). For one protocol, eight measurements were made to assess the variation. All measurements were made using a test mode heart rate of 60 bpm (Table 3). Table 3 shows a higher variation in dose for the right breast when

compared with the left. This may be because separate MOSFET dosimetry systems were used in the right and left breasts (Figure 8b). Previous performance measurements of the two MOSFET systems had shown a higher measurement variation with constant dose for the system used in the right breast than the left.

Lifetime attributable risk of cancer

The MOSFET dosimetry measurements were compared with the ImPACT dosimetry calculator. Mean breast doses (averaged for both breasts) were extrapolated over a clinically feasible mA range (0–1000 mA,

Figure 7. Variation in entrance surface dose (ESD) (single axial rotation) for cardiac medium, cardiac small and cardiac large bow tie filters on GE HD750 CT scanner (General Electrical Healthcare, Milwaukee, WI). Note that the radiation start/stop locations appear random. Increased noise at higher doses associated with calibration (power relationship). Asymmetry possibly explained by primary beam overlap incident on cylindrical phantom (not present for cardiac small filter owing to reduced field of view compared with cardiac medium and large filters).

Figure 8. (a) Position of the metal oxide semiconductor field-effect transistor detectors in breast phantom. (b) Position of atom phantom within scanner bore. Note ECG test waveform at 60 bmp.

corresponding to 0–227 mA per rotation). A linear relationship of dose with mA was assumed when extrapolating the 500 mA MOSFET measurement. The results (Figure 9) show a good correlation between absorbed doses measured with the MOSFET detectors and the ImPACT calculator. The combined tissue-weighted equivalent dose to the breasts and the lung was found to account for more than two-thirds of the effective dose for CTCA (Table 4). It is estimated that typical equivalent breast doses for clinical prospectively gated CTCA protocols (GE HD750) at our institution range between 2 mSv and 15 mSv. Lifetime attributable risk (LAR) of cancer was calculated for prospective CTCA protocol based on organ doses estimated from the ImPACT calculator. LAR is shown for females (breast and lung) and males (lung only) for the ages of 20 years and 50 years at irradiation. For a typical prospective protocol (GE HD750, cardiac medium, 120 kV, 400 mA) at this institution, the LAR for solid cancer incidence of the breast of a 20-year-old female patient is 0.05%

(51 cases in 100 000 exposed). For interest, breast dose was measured for a retrospective protocol (SnapShot Burst (General Electric Healthcare), cardiac medium bow tie filter, 120 kV, 400 mA, 0.2:1 overlapping helical pitch) using the MOSFET technique. A mean absorbed breast dose of ~80 mGy was measured (80 mSv equivalent dose) representing a LAR solid cancer incidence of 0.34% for a 20-year-old female.

We have recently published work [8] comparing effective radiation doses between IA and prospectively gated CTCA. To enable LAR comparison for breast cancer incidence between IA and CTCA the raw data from this audit was re-analysed within a PC-based X-ray Monte Carlo model programme PCXMC version 2.20 [21]. The PCXMC simulator is based on the Cristy and Eckerman model and incorporates the new ICRP 103 tissue weighting factors. For a BMI restricted range of 20–35, the mean equivalent dose to the female breast for IA was 1.3 ± 0.4 mSv ($\pm$ standard deviation) corresponding to a LAR of 6 per 100 000 exposed when IA is undertaken at

Table 3. Table shows high variation in absorbed dose recorded in the centre plane of the Atom phantom breasts for eight successive measurements. Scan protocol: SnapShot pulse™ (General Electric Healthcare, Milwaukee, WI), 120 kV cardiac small bow tie filter, 500 mA (113.5 mAs) 4 slabs (140 mm coverage)

Breast ID	MOSFET location	Measurement number								Mean	SD
		1	2	3	4	5	6	7	8		
Right	Centre	2.94	3.84	6.29	13.40	20.50	14.00	14.40	4.30	9.96	6.45
	Head	19.30	17.60	7.66	12.30	4.88	5.36	19.80	2.68	11.20	6.98
	Foot	18.00	11.80	16.80	15.60	2.02	12.60	21.30	15.10	14.15	5.74
	Left	15.50	19.10	10.80	14.00	17.80	14.70	1.20	10.40	12.94	5.62
	Right	2.81	9.80	6.17	5.79	14.20	11.40	15.40	9.22	9.35	4.31
	Mean	11.71	12.43	9.54	12.22	11.88	11.61	14.42	8.34	11.52	5.82
Left	Centre	9.50	14.90	16.90	7.29	16.50	14.10	13.50	14.60	13.41	3.35
	Head	14.40	8.48	13.70	15.40	7.01	12.30	13.70	13.50	12.31	2.98
	Foot	11.18	5.87	6.78	16.70	11.70	17.60	17.60	6.15	11.70	5.13
	Left	12.10	15.20	14.60	6.49	16.80	13.30	8.75	15.10	12.79	3.53
	Right	9.70	12.60	12.00	8.51	12.60	7.67	7.13	12.80	10.38	2.40
	Mean	11.38	11.41	12.80	10.88	12.92	12.99	12.14	12.43	12.12	3.48

MOSFET, metal oxide semiconductor field-effect transistor; SD, standard deviation.
All doses in mGy.

Figure 9. Comparison of absorbed breast dose measured with metal oxide semiconductor field-effect transistor (MOSFET) detectors and calculation by imaging performance assessment of CT scanners (imPACT) dosimetry calculator for various kilovoltage/bow tie filters over a clinically feasible milliamp range. All for prospective CT coronary angiography protocol. MOSFET results extrapolated below and above tube current of 500 mA.

the age of 20 years. When IA is performed at the age of 50 years, this falls to 1 in 100 000. The LAR for breast cancer incidence for IA is therefore approximately 9 times less than for prospectively gated CTCA at this institution (when compared with typical protocol of cardiac medium bow tie filter, 120 kV, 400 mA tube current). This is expected as, with the majority of X-ray projections during IA performed with the X-ray tube under the X-ray table with the patient lying supine, the body significantly attenuates the primary radiation field before reaching the breasts.

Discussion and conclusions

This work demonstrated that it is possible to oversimplify the methods used in the estimation of effective dose for CTCA. When using prospective gating, consideration has been given to the effect partial irradiation has in the calculation of both the effective dose and absorbed dose to the breast. An adapted method for measuring $CTDI_w$ for segmental CT irradiation had been developed and used to calculate scanner specific effective doses for CTCA [8].

Radiochromatic film has been utilised to evaluate the magnitude and distribution of ESD for prospectively gated CTCA techniques. We have shown that when considering a single segmented axial rotation for prospectively gated CTCA, the ESD to the skin surface may vary from 2 to 16 mGy for our institution-specific clinical protocols. When averaged for a typical 4-slab scan (to achieve a typical cardiac scan length of 140 mm) the result is a mean ESD of approximately 8 mGy with significantly less variation around the

body's circumference. When considering the organ dose to the breast from such fixed mA prospective protocols on a patient-by-patient basis, there is likely to be some variation as a function of random start/stop locations for the partial axial irradiation. For population doses (effective dose/ESD/organ dose), this effect is diluted and these methods should allow meaningful comparison with other radiographic modalities.

MOSFET detectors have been used to estimate organ dose to the breast for an average sized patient protocol under prospective gating. Whilst recorded doses were observed to vary on a scan-by-scan basis depending on the seemingly random location of segmental irradiation, equivalent breast doses of approximately 7–20 mSv were observed. This compares favourably (8–15 mSv) with similar work by Einstein et al [20] using MOSFET detectors exposed within an identical phantom on a 320-detector row CT scanner (Aquilion ONE; Toshiba Medical Systems, Otawara, Japan). For comparison, the average breast dose for an oblique projection mammogram in the National Health Service breast screening program is 2.03 ± 0.08 mGy [22] (mean glandular dose for the typical 50–60 mm compressed breast thickness).

The LAR of solid cancer incidence to the breast for a 20-year-old female patient (exposed to a typical prospective CTCA protocol) was found to be $\sim 0.05\%$; this risk falls to $\sim 0.008\%$ for a 50-year-old female exposed to the same protocol. For interest, the breast dose for a retrospectively gated protocol was measured. Breast doses for this protocol were found to be approaching 100 mGy. Whilst breast doses for a typical prospectively gated protocol were found to be significantly lower, mean breast doses from IA at this institution were found to be approximately nine times less than prospectively gated CTCA. We have

Table 4. Lung, breast and effective doses for CT coronary angiography (CTCA) for various fixed mA prospective CTCA protocols. All organ doses calculated from use of ImPACT dosimetry calculator. Lifetime attributable risk (LAR) calculated from annex 12D of BEIR VII phase 2 [7]. The LAR 20/50 is the statistical incidence of breast cancer (for a 20-year-old and 50-year-old patient, respectively) per 100 000 people exposed to the stated organ dose

Filter: cardiac small, 100 kV													
mA	Absorbed dose breast (mGy)	Tissue-weighted equivalent dose breast (mSv)	Absorbed dose lung (mGy)	Tissue-weighted equivalent dose lung (mSv)	Effective dose ImPACT	%Effective dose breast plus lung	Female lifetime attributable risk of cancer incidence (per 100 000 exposed)				Male LAR/ 100 000		
							LAR 20 years breast	LAR 50 years breast	LAR 20 years lung	LAR 50 years lung	LAR 20 years lung	LAR 50 years lung	DLP (mGy x cm)
100	1.3	0.16	1.2	0.14	0.45	67	6	1	4	3	2	1	18
200	2.5	0.30	2.4	0.29	0.91	65	11	2	6	6	4	2	37
300	3.8	0.46	3.7	0.44	1.4	64	16	3	13	9	6	4	55
400	5.1	0.61	4.9	0.59	1.8	67	22	4	17	11	7	5	74
500	6.3	0.76	6.1	0.73	2.3	65	27	4	21	14	9	6	92
600	7.6	0.91	7.3	0.88	2.7	66	33	5	25	17	11	7	111
700	8.9	1.07	8.5	1.02	3.2	65	38	6	29	20	13	9	129
800	10	1.20	9.8	1.18	3.6	66	43	7	34	23	15	10	148
900	11	1.32	11	1.32	4.1	64	47	8	38	25	16	11	166
1000	13	1.56	12	1.44	4.5	67	56	9	42	28	18	12	185
Filter: cardiac small, 120 kV													
mA	Absorbed dose breast (mGy)	Tissue-weighted equivalent dose breast (mSv)	Absorbed dose lung (mGy)	Tissue-weighted equivalent dose lung (mSv)	Effective dose ImPACT	%Effective dose breast plus lung	Female lifetime attributable risk of cancer incidence (per 100 000 exposed)				Male LAR/ 100 000		
							LAR 20 years breast	LAR 50 years breast	LAR 20 years lung	LAR 50 years lung	LAR 20 years lung	LAR 50 years lung	DLP (mGy x cm)
100	2.1	0.25	2	0.24	0.75	66	9	1	7	5	3	2	30
200	4.3	0.52	4	0.48	1.5	66	18	3	14	9	6	4	61
300	6.4	0.77	6	0.72	2.3	65	27	4	21	14	9	6	91
400	8.6	1.03	8	0.96	3	66	37	6	28	18	12	8	122
500	11	1.32	10	1.20	3.8	66	47	8	35	23	15	10	152
600	13	1.56	12	1.44	4.5	67	56	9	42	28	18	12	182
700	15	1.80	14	1.68	5.3	66	64	11	48	32	21	14	213
800	17	2.04	16	1.92	6	66	73	12	55	37	24	16	243
900	19	2.28	18	2.16	6.8	65	82	13	62	41	27	18	274
1000	21	2.52	20	2.40	7.5	66	90	15	69	46	30	20	304

Table 4. Continued

Filter: cardiac medium, 120 kV														
mA	Absorbed dose breast (mGy)	Tissue-weighted dose breast (mSv)	Absorbed dose lung (mGy)	Tissue-weighted dose lung (mSv)	Effective dose ImPACT	%Effective dose breast plus lung	Female lifetime attributable risk of cancer incidence (per 100 000 exposed)					Male LAR/100 000		
							LAR 20 years breast	LAR 50 years breast	LAR 20 years lung	LAR 50 years lung	LAR 20 years lung	LAR 50 years lung	LAR 20 years lung	LAR 50 years lung
100	3	0.36	2.1	0.25	0.83	74	13	2	7	5	3	2	34	
200	6	0.72	4.1	0.49	1.7	71	26	4	14	9	6	4	69	
300	8.9	1.07	6.2	0.74	2.5	72	38	6	21	14	9	6	103	
400	12	1.44	8.2	0.98	3.3	73	51	8	28	19	12	8	137	
500	15	1.80	10	1.20	4.1	73	64	11	35	23	15	10	172	
600	18	2.16	12	1.44	5	72	77	13	42	28	18	12	206	
700	21	2.52	14	1.68	5.8	72	90	15	48	32	21	14	241	
800	24	2.88	16	1.92	6.6	73	103	17	55	37	24	16	275	
900	27	3.24	19	2.28	7.5	74	116	19	66	44	28	19	309	
1000	30	3.6	21	2.52	8.3	74	129	21	73	48	31	21	344	

Filter: cardiac large, 140 kV														
mA	Absorbed dose breast (mGy)	Tissue-weighted dose breast (mSv)	Absorbed dose lung (mGy)	Tissue-weighted dose lung (mSv)	Effective dose ImPACT	%Effective dose breast plus lung	Female lifetime attributable risk of cancer incidence (per 100 000 exposed)					Male LAR/100 000		
							LAR 20 years breast	LAR 50 years breast	LAR 20 years lung	LAR 50 years lung	LAR 20 years lung	LAR 50 years lung	LAR 20 years lung	LAR 50 years lung
100	3.6	0.43	3	0.36	1.1	72	15	3	10	7	4	3	42	
200	7.1	0.85	6	0.72	2.3	68	30	5	21	14	9	6	85	
300	11	1.32	9	1.09	3.4	71	47	8	31	21	14	9	127	
400	14	1.68	12	1.44	4.6	68	60	10	42	28	18	12	170	
500	18	2.16	15	1.80	5.7	69	77	13	52	35	22	15	212	
600	21	2.52	18	2.16	6.9	68	90	15	62	41	27	18	254	
700	25	3.00	21	2.52	8	69	107	18	73	48	31	21	297	
800	29	3.48	24	2.88	9.2	69	124	20	83	55	36	24	339	
900	32	3.84	27	3.24	10	71	137	22	93	62	40	27	382	
1000	36	4.32	30	3.60	11	72	154	25	104	69	45	30	424	

shown that the sum of the equivalent doses to the breast and lung for CTCA accounts for two-thirds of the total effective dose to the patient. Significant reductions in effective dose for CTCA could therefore be realised by protecting the breast tissue during CT examination (either by limiting the X-ray tube exposure time to segments covering the patients back, utilising physical breast shields or simply reducing exposure factors). As ever there is always a trade-off between technological capability, image quality and dose that will limit dose reduction strategy. Optimisation of cardiac protocols (for example the increased use of low kilovoltage protocols and further introduction of iterative reconstruction) is expected to show further reduction in dose.

Acknowledgment

All practical work was undertaken at Plymouth Hospitals NHS Trust.

Conflicts of interest

R J Loader, C Roobottom and G Morgan-Hughes have all received lecture fees from General Electric Healthcare.

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