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ORIGINAL PAPER

Radiation dose values for various coronary calcium scoring protocols in dual-source CT

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Abstract *Purpose* The purpose of this study was to assess the radiation dose and associated image noise of previously suggested calcium scoring protocols using dual-source CT. *Methods* One hundred consecutive patients underwent coronary calcium scoring using dual-source CT. Patients were randomly assigned to five different protocols: retrospective ECG-gating and tube current reduction to 4% outside the pulsing window at 120 (protocol A) and 100 kV (B), prospective ECG-triggering at 120 (C) and 100 kV (D), and prospective ECG-triggering at 100 kV with attenuation-based tube current modulation (E). Radiation dose parameters and image noise were determined and compared. *Results* Protocol A resulted in an effective dose of 1.3 ± 0.2 mSv, protocol B in 0.8 ± 0.2 mSv, protocol C in 1.0 ± 0.2 mSv, protocol D in 0.6 ± 0.1 mSv, and protocol E in 0.7 ± 0.1 mSv. Effective doses were significantly lower ($P < 0.001$) with 100 kV when compared to 120 kV protocols, and were significantly lower ($P < 0.001$) for prospective versus retrospective ECG-gating. No significant difference was found between protocol D and E. Significant negative correlations were found between the $CTDI_{vol}$ and heart rate for both retrospective ECG-gating protocols (protocol A: $r = -0.98$, $P < 0.001$; protocol B: $r = -0.83$, $P < 0.001$). The mean image noise was 29.0 ± 6.7 HU, with no significant differences between the five protocols. The image noise was significantly correlated with the body weight ($r = 0.21$, $P < 0.05$) and BMI ($r = 0.31$, $P < 0.01$). *Conclusions* Effective dose of calcium scoring using dual-source CT ranges from 0.6 to 1.3 mSv. Prospective triggering and lower tube voltage significantly reduces the radiation but yield similar image noise.

Keywords Coronary calcium scoring · Dual-source computed tomography · Radiation dose

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Introduction

Coronary calcium scoring (CS) using computed tomography (CT) has been validated as a tool for optimizing risk stratification regarding the development of non-fatal and fatal cardiac events [1–5]. Recent guidelines from the American Heart Association [6] endorsed the screening using CS as a method to reclassify risk in patients at intermediate risk based on traditional scores such as the Framingham and Procam algorithms.

CS, being a screening tool, requires the use of low radiation dose techniques in order to outweigh the potential risks of the examination [6], while at the same time maintain a reasonable image quality, reliability and accuracy [7]. In order to keep radiation exposure as low as reasonably achievable (ALARA), recent modifications of multi-slice CT scanning protocols have been implemented, such as the use of a low tube current [8, 9], attenuation-based tube current modulation [10], and use of low tube voltage protocols [11, 12]. In general, two modes of phase synchronization have been traditionally used, i.e., the retrospective electrocardiography (ECG)-gating technique [13, 14] that acquires continuous data in a helical (spiral) mode and the prospective ECG-triggering technique [15, 16] that obtains data at predefined time points of the cardiac cycle in an axial step-and-shoot mode, the latter usually being associated with a lower radiation dose. Radiation dose can also be significantly reduced in ECG-gated helical (spiral) examinations by using ECG-controlled modulation of the output of the X-ray [17].

The dual-source CT system has been recently introduced which, by virtue of its tube and detector configuration, is characterized by a high and heart rate independent temporal resolution [18]. This allows cardiac scanning of diagnostic image quality even at high and irregular heart rates [19]. In an ECG-gated helical mode, dual-source CT allows the flexible adjustment of the ECG-pulsing window to the individual heart rate [20] and, through adaptation of the spiral pitch to the patient's heart rate, is characterized by decreasing radiation dose at higher heart rates [21, 22].

The purpose of this study was to assess the radiation dose and associated image noise of previously suggested CS protocols using dual-source CT.

Methods

Study population

One hundred consecutive, asymptomatic patients (38 females, 62 males, mean age 69 ± 11 years, range 44–88 years) were included in this study. All were referred to our department for CS. Personal data including body weight and height were recorded. The body mass index (BMI) was calculated from body weight and height. Patient characteristics are shown in Table 1. Patients with irregular heart rates were not excluded from this study. The study had approval from our local ethical committee who waived the written informed consent requirement.

Dual-source CT calcium scoring protocol

All CS examinations were performed on a dual-source CT scanner (Somatom Definition, Siemens Medical Solutions, Forchheim, Germany) using the following scan parameters: detector collimation $2 \times 32 \times 0.6$ mm, slice acquisition $2 \times 64 \times 0.6$ mm by

Table 1 Patient characteristics ($n = 60$)

	Protocol A	Protocol B	Protocol C	Protocol D	Protocol E	*P*-Value
Female (%)	6 (30)	6 (30)	10 (59)	5 (25)	11 (55)	0.19
Male (%)	14 (70)	14 (70)	10 (50)	15 (75)	9 (45)	0.19
Age (yrs)	68.6 ± 12.4	69.7 ± 12.4	70.6 ± 11.5	69.9 ± 10.3	65.3 ± 9.9	0.55
BMI (kg/m^2)	26.2 ± 3.3	27.0 ± 4.0	25.2 ± 4.0	25.8 ± 2.2	24.8 ± 3.8	0.23
Heart rate (bpm)	72.2 ± 11.8	69.4 ± 14.2	66.9 ± 15.1	62.5 ± 9.4	68.7 ± 12.1	0.14
Nicotine abuse (%)	7 (35)	5 (25)	9 (45)	6 (30)	9 (45)	0.60
Hypertension (%)	8 (40)	6 (30)	8 (40)	4 (20)	10 (50)	0.34
Hypercholesterolemia (%)	5 (25)	6 (30)	6 (30)	3 (15)	6 (30)	0.78
Diabetes mellitus (%)	3 (15)	2 (10)	4 (20)	2 (10)	2 (10)	0.84
Family history (%)	3 (15)	5 (25)	5 (25)	7 (35)	2 (10)	0.35
Agatston score	380 ± 593	355 ± 513	266 ± 289	215 ± 402	299 ± 527	0.55

Data are presented as rates and frequencies or means ± SD

Chi-square and Kruskal–Wallis tests were used to assess for significant differences means of a z-flying focal spot, and gantry rotation time 330 ms. No beta-receptor antagonists for heart rate control were administered prior to CT. The region imaged extended from the level of the aortic root to the diaphragm.

Acquisition techniques and protocols

All 100 patients were randomly assigned to one of the following five CS protocols that were modeled after previously suggested protocols:

Protocol A Retrospective ECG-gating (i.e., helical mode) with ECG-pulsing at 70% of the RR-interval and a reduction of tube current to 4% outside a reconstruction window (minimum width starting at 70% of the RR-interval) using a tube voltage of 120 kV and a tube current time product of 80 mAs per rotation ($n = 20$), in accordance to the literature [10, 23].

Protocol B Retrospective ECG-gating (i.e., helical mode) with ECG-pulsing at 70% of the RR-interval and a reduction of tube current to 4% outside a reconstruction window (minimum width starting at 70% of the RR-interval) using a tube voltage of 100 kV and a tube current time product of 80 mAs per rotation ($n = 20$), in accordance to the literature [12].

Protocol C Prospective ECG-triggering (i.e., axial step-and-shoot mode) with a cycle time of 1.36 s and using a tube voltage of 120 kV and a tube-current time product of 80 mAs per rotation ($n = 20$), in accordance to the literature [15].

Protocol D Prospective ECG-triggering (i.e., sequential mode) with a cycle time of 1.36 s and using a tube voltage of 100 kV and a tube current time product of 80 mAs per rotation ($n = 20$), in accordance to the literature [15].

Protocol E Prospective ECG-triggering (i.e., sequential mode) with a cycle time of 1.36 s and using a tube voltage of 100 kV and an attenuation-based tube current modulation (reference tube current time product set at 80 mAs) ($n = 20$), in accordance to the literature [24].

With the use of an attenuation-based tube current time modulation, the reference tube current time product is modulated in the *z*-axis direction following the patients' attenuations. When implemented in the cardiac scan template, the tube current is calculated based on the mean attenuation values derived from the scanogram and set constant for all *z*-axis positions.

Image reconstruction and data evaluations

All images were reconstructed with a mono-segment reconstruction algorithm [18]. CT data were reconstructed at 70% of the RR-interval using a slice thickness of 3 mm and an increment of 3 mm. All data were transferred to an external workstation; calcifications were quantified with cardiac post-processing software (Syngo CaScore, Siemens) by one experienced observer with 4 years of experience in cardiovascular radiology.

Radiation dose

Dual-source CT behaves similar to conventional CT dose metrics since radiation from both single tubes sum up in a linear manner [21].

The CT volume dose index ($CTDI_{vol}$), being one fundamental radiation dose parameter in CT averages radiation dose over the centre slice of a CT examination consisting of multiple parallel slices [25–27]. Thereby, the numeric value of the $CTDI_{vol}$ is directly related to the degree of overlap between adjacent slices which is determined by the width of the individual slices and by their distance. In the helical scanning mode, the distance between adjacent slices is dependent on how far the patient table advances during one gantry rotation [26]. This so called helical pitch was recorded in all examinations that were done in the helical mode. Scan lengths were noted in all patients. The $CTDI_{vol}$ when using ECG-based tube current modulation is proportional to the average tube current of a single acquisition cycle [21].

The dose-length product (DLP) represents the integrated radiation dose imparted by all slices of a CT examination. The DLP is defined by the scan length, multiplied by the $CTDI_{vol}$ [26].

The parameter of effective dose is an estimate of the dose to patients during an ionizing radiation procedure. It measures the total energy entered into the body and then takes into account the sensitivity of the organs irradiated [25]. The effective dose also allows direct comparison with other sources of radiation exposure, and is the preferred clinical measure of exposure with CT. The effective dose

($Dose_{eff}$) was calculated as previously recommended and validated [22, 25, 26, 28].

Image noise

Objective image quality of the five protocols was determined by a radiologist with 4 years of experience in cardiac radiology. To measure image noise, attenuation measures were performed applying a standardized circular region of interest in the ascending aorta at the level of the origin of the coronary arteries on two consecutive axial slices (Fig. 1). The region of interest was defined as large as possible (mean 2.9 ± 1.4 cm^2, range 1.8–4.3 cm^2) carefully avoiding the vessel wall or plaques. The standard deviations (SD) of the attenuation measurements within the ascending aorta were ascribed to image noise. The mean value of these two measurements was calculated for each patient.

The influence of image noise on CS was assessed. A radiologist (with 2 years of experience in cardiac radiology) who was blinded to the results from objective image quality assessment rated the coronary arteries on a 2-point scale: score 1, image noise adequate for CS; no non-calcified voxels above threshold; score 2, image noise potentially inadequate for scoring; more than one non-calcified voxels above threshold.

Statistical analysis

Categorical variables are expressed as frequencies or percentages. Numerical values of continuous variables are expressed as means and SD.

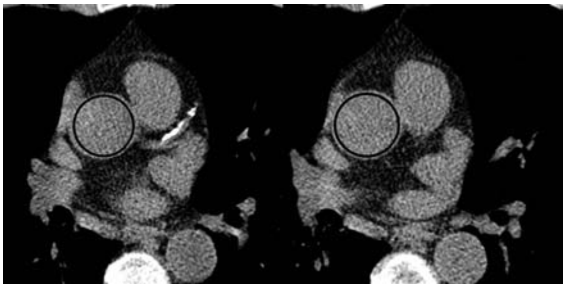

Fig. 1 Image noise measurements. The region of interest was placed into the ascending aorta on two consecutive axial slices and defined as large as possible avoiding partial volume effects. Standard deviations of the attenuation measurements within the ascending aorta were ascribed to image noise. The mean value of both measurements was taken for analysis

Chi-square and Kruskal–Wallis tests were used to assess demographics (i.e., cardiovascular risk factors, heart rate, age, BMI, and Agatston score, as well as scanning range) for significant differences. A *P*-value of less than 0.05 was considered to indicate statistical significance. Pairwise Mann–Whitney-*U* tests of the specific protocols were used to ascertain significance of differences in regard to dose parameters (i.e., $CTDI_{vol}$, DLP, and $Dose_{eff}$) and image noise. Image noise was compared for normal (BMI $\leq$ 25 kg/m^2) and overweight patients (BMI $>$ 25 kg/m^2) using the Kruskal–Wallis tests. The Kruskal–Wallis test was also used to assess differences between the protocols in the rate of inadequate image noise. According to the Bonferroni method, the level of confidence ($P = 0.05$) was corrected for multiple comparisons between the different protocols. For continuous data (i.e., heart rate, body weight, BMI, $CTDI_{vol}$, image noise) correlation analysis was performed using Spearman rank order correlation coefficients. Data analysis was performed using commercially available software (SPSS 12.0, Chicago, USA).

Results

All CT examinations were performed without complications. The mean BMI in the 100 patients was 25.2 ± 3.7 kg/m^2 (range 17.3–35.2 kg/m^2). The mean heart rate during scanning was 67 ± 13 bpm (range 43–106 bpm), and the mean scanning range was 128 ± 20 mm (range 95–149 mm), with no statistical differences ($P = 0.94$) among the five protocol protocols (Table 2).

Radiation dose

The radiation dose parameters for all protocols are shown in Table 2.

Significant differences ($P < 0.001$) were found between protocol A (i.e., helical 120 kV) and protocol B (i.e., helical 100 kV) for $CTDI_{vol}$, DLP, and $Dose_{eff}$. Similarly, significantly differences ($P < 0.001$) were revealed for $CTDI_{vol}$, DLP, and $Dose_{eff}$ values when protocol C (sequential 120 kV) was compared to protocol D (i.e., sequential 100 kV).

We found significant differences ($P < 0.001$) between protocol A (i.e., helical 120 kV) and C (sequential 120 kV) for $CTDI_{vol}$, DLP, as well as for

Table 2 Radiation dose parameters of different scanning protocols

	Retrospective ECG-gating		Prospective ECG-triggering		
	Protocol A	Protocol B	Protocol C	Protocol D	Protocol E
Scan length	12.9 ± 1.1	12.7 ± 1.5	12.8 ± 1.2	12.9 ± 1.2	12.5 ± 0.7
$CTDI_{vol}$ [mGy]	5.85 ± 0.77	3.71 ± 0.71	4.54 ± 0.01	2.62 ± 0.02	3.10 ± 0.71
DLP [mGy/cm]	75.3 ± 11.7	47.4 ± 12.1	58.3 ± 11.3	33.7 ± 6.5	38.8 ± 7.6
$Dose_{eff}$ [mSv]	1.3 ± 0.2	0.8 ± 0.2	1.0 ± 0.2	0.6 ± 0.1	0.7 ± 0.1

ECG Electrocardiography, *$CTDI_{vol}$* CT volume dose index, *DLP* Dose-length product, *$Dose_{eff}$* Effective dose

Data are means ± SD. Scan length was not significantly different between the five different protocols

$Dose_{eff}$. Protocol B (i.e., helical 100 kV) had significantly ($P < 0.001$) higher $CTDI_{vol}$, DLP, and $Dose_{eff}$ values when compared to protocol D (i.e., sequential 100 kV).

No significant differences were found when comparing protocol D (i.e., sequential 100 kV) and protocol E (i.e., sequential 100 kV with attenuation-based tube current modulation) regarding $CTDI_{vol}$ ($P = 0.16$), DLP ($P = 0.27$), and $Dose_{eff}$ ($P = 0.27$).

A significant negative correlation was found between $CTDI_{vol}$ and heart rate in protocol A ($r = -0.98$, $P < 0.001$) and B ($r = -0.83$, $P < 0.001$). Scatterplots and correlations are shown in Fig. 2a. No significant correlation was found between $CTDI_{vol}$ and heart rate in protocol C, D, E ($P > 0.05$, Fig. 2b).

$CTDI_{vol}$ values in protocol E (i.e., sequential 100 kV with attenuation-based tube current modulation) showed a significant correlation with the patients body weight ($r = 0.63$, $P < 0.05$) and BMI ($r = 0.84$, $P < 0.01$).

Image noise

In regard to protocols utilizing retrospective ECG-gating, the mean image noise was 28.9 ± 6.4 HU (range 21.0–44.0 HU) in protocol A, and was 32.3 ± 8.1 HU (range 19.0–48.0 HU) in protocol B, respectively (Table 3).

In regard to protocols utilizing prospective ECG-triggering, the mean image noise was 27.2 ± 5.6 HU (range 18.0–39.0 HU) in protocol C, and was 29.3 ± 6.6 HU (range 21.0–44.0 HU) in protocol D, respectively. The mean image noise of protocol E using an attenuation-based tube current modulation was 32.3 ± 5.8 HU (range 22.0–48.0 HU).

In regard to protocols utilizing retrospective ECG-gating, no significant difference ($P = 0.18$) were found between protocol A (i.e., helical 120 kV) and protocol B (i.e., helical 100 kV) for image noise. Similarly in regard to protocols utilizing prospective

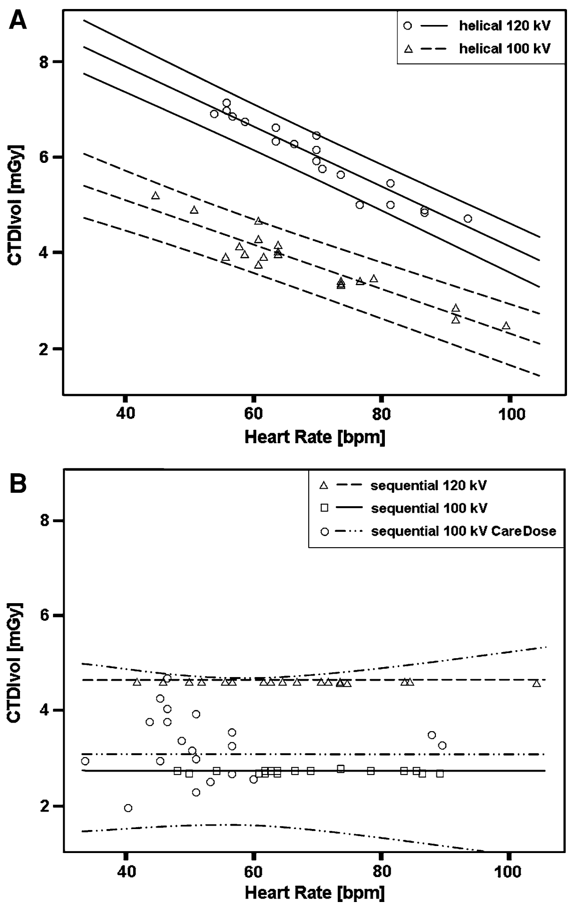

Fig. 2 Scatterplots and correlation between $CTDI_{vol}$ [mGy] and heart rate [bpm] for the different protocols. **a** A significant negative correlation was found for the helical protocol at both tube voltages ($r = -0.98$ and $r = -0.83$, $P < 0.001$, respectively). **b** No significant correlation was found for all sequential protocols ($P =$ n.s.)

Table 3 Image noise of the different calcium scoring protocols

	Image noise	Protocol A	Protocol B	Protocol C	Protocol D	Protocol E
Protocol A	28.9 ± 6.4	–	0.18	0.60	0.93	0.12
Protocol B	32.3 ± 8.1	n.s.	–	0.06	0.23	0.86
Protocol C	27.2 ± 5.6	n.s.	n.s.	–	0.48	0.05
Protocol D	29.3 ± 6.6	n.s.	n.s.	n.s.	–	0.06
Protocol E	32.3 ± 5.8	n.s.	n.s.	n.s.	n.s.	–

Data are presented as means ± standard deviation. The Mann–Whitney-U test was used to test for the significance of differences

ECG-triggering, image noise did not differ significantly ($P = 0.48$) when protocol C (sequential 120 kV) was compared to protocol D (i.e., sequential 100 kV).

We did not find significant differences between protocol A (i.e., helical 120 kV) and C (sequential 120 kV) ($P = 0.60$), and between protocol B (i.e., helical 100 kV) had and protocol D (i.e., sequential 100 kV) ($P = 0.23$). Because the mean image noise of all protocols was comparable (Fig. 3), the mean image noise was calculated with 29.0 ± 6.7 HU.

No significant differences were found when comparing protocol D (i.e., sequential 100 kV) and protocol E (i.e., sequential 100 kV with attenuation-based tube current modulation) ($P = 0.06$).

No significant differences were found when comparing the image noise in normal weight patients between the protocols ($P = 0.17$). In regard to image noise in overweight patients, significant differences where found ($P < 0.05$), with a higher mean image noise using 100 kV protocols.

The image noise of all patients was significantly correlated with the body weight ($r = 0.21$, $P < 0.05$) and BMI ($r = 0.31$, $P < 0.01$). When correlating the image noise with the body weight and BMI for the protocols separately, no level of significance was found for any protocol except of protocol B (body weight, $r = 0.51$, $P < 0.05$; BMI, $r = 0.50$, $P < 0.05$).

In regard to protocols utilizing retrospective ECG-gating, coronary arteries were rated as having a score 1 in 20/20 patients (100%) using protocol A and in 18/20 patients (90%) with protocol B.

In regard to protocols utilizing prospective ECG-triggering, coronary arteries were rated as having a score 1 in 19/20 segments (95%) using protocol C, in 18/20 segments (90%) with protocol D, and in 17/20 patients (85%) with protocol E, respectively. Non-calcified voxels above the threshold were exclusively present in the posterior descending artery in 5/8 patients (63%). The rate of coronary arteries with inadequate image noise (score 2) was not significantly different ($P > 0.05$) among the five different protocols.

Discussion

This study shows that the effective radiation dose for coronary CS using dual-source CT ranges between 0.6 and 1.3 mSv, depending on the protocol used. Prospective triggering significantly reduced the radiation dose when compared to the retrospectively ECG-gated protocols when using a similar tube voltage. The $CTDI_{vol}$ showed a significant negative correlation with increasing heart rates while using the retrospective ECG-gating protocol. The reduction of

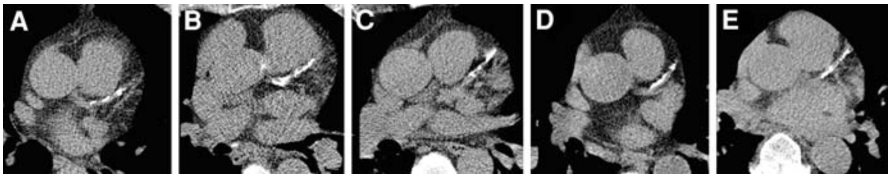

Fig. 3 Image examples of the five protocols in five different male patients having a body mass index between 27.3 and 28.2 kg/m^2. Coronary arteries were considered to be depicted with adequate image noise for calcium scoring. **a** Helical 120 kV. **b** Helical 100 kV. **c** Sequential 120 Kv. **d** Sequential 100 kV. **e** Sequential 100 kV with attenuation-based tube current modulation

the tube potential was associated with a significant reduction in radiation dose for both phase synchronizing techniques. Despite of the various dose characteristics, all protocols yielded similar image noise levels.

Although the usefulness of CS with regard to risk stratification has been proven in a number of studies [1–3], the optimal CT scanning protocols are still a matter of discussion [6, 7, 29–33]. Some authors have demonstrated the best reproducibility for the retrospective ECG-gating technique [13, 15, 30]. This becomes important when tracking changes over time in order to assess the efficacy of therapies [6]. Because protocols using retrospective ECG-gating are generally associated with a higher dose as compared to protocols employing the prospective technique, various efforts have been made for developing and optimizing dose saving algorithms for retrospective ECG-gated CT [12, 17, 18, 34, 35].

The heart rate adapted pitch of dual-source CT has been previously shown to be the major contributor to minimized dose at higher heart rates [21, 22]. The $CTDI_{vol}$ in our study significantly correlated with the heart rate for both retrospectively ECG-gated protocols. In a recent CT coronary angiography study of Stolzmann et al. [22], the authors have found no significant correlation between $CTDI_{vol}$ and heart rate when using the 4% tube current reduction protocol. In contrast, this study shows a significant correlation between $CTDI_{vol}$ and heart rate that can be explained by the fact that for CS, in contrast to CT coronary angiography, the ECG-pulsing window can be kept narrow also at higher heart rates.

It has been shown that lower tube voltages considerably reduce the radiation dose of CT examinations [11, 36, 37]. For CS, this relevant dose reduction is possible without compromising the reproducibility and accuracy of the method. At lower tube voltages, however, the detection threshold needs to be increased because of the increasing attenuation of calcium at lower photon energy levels [37].

With a reduction of tube potential from 120 kV to 100 kV, it is possible to save approximately 37% of the delivered radiation dose when using retrospective ECG-gating and approximately 40% when using the prospective ECG-triggering technique.

Recently, Muhlenbruch et al. [10] showed that use of attenuation-based tube current for CS leads to a more balanced image noise and reduces the radiation dose as compared to a fixed tube current. By doing so, the dose usage can be optimized without overexposure in low weighted or underexposure in overweight patients. In our study, the use of attenuation-based tube current resulted in a $CTDI_{vol}$ that significantly correlated with the weight and BMI of the patients, as expected. As compared to the fixed tube current protocol, a slight but non-significant increase of the $CTDI_{vol}$ was noted that is most likely explained by our population being on average overweight.

Interestingly in our study, besides significant reductions radiation dose were found, all protocols yielded similar image noise. However, image noise differed among the five protocols when only considering overweight patients. In contrast to Mahnken et al. [38] who described a mean image noise level of 19 HU, we observed a slightly higher mean image noise of 29.0 HU. The main reason for this is that the tube voltage in the study by Mahnken et al. [38] was set to 140 kV whereas in our study all scans were performed at either 100 or 120 kV in this overweight patient cohort leading to an increased image noise. However, the measured mean image noise levels were close to the image noise recommendation for coronary calcium scoring of 25 HU [10].

Study limitations

First, we have not assessed the accuracy and reproducibility of the various coronary CS protocol tested in this study. However, this was not the purpose of this study which rather aimed to determine radiation dose parameters associated with the various protocols. Certainly, future studies must aim at an evaluation of the most accurate and robust dual-source CT CS protocol being associated with the lowest possible radiation dose [7]. Second, we used in most protocols a fixed tube current time product of 80 mAs per rotation as recommended by the vendor. This might not reflect the optimal tube current with regard to accuracy and radiation dose, which let appear our dose calculations being relatively arbitrary. A recent phantom study on CS has shown that size-specific tube current values yielded comparable data among different patient sizes and thus recommended the implementation of adjusted tube current time product settings for small, medium and large patients [9]. This is in line with the results from the study of Muhlenbruch et al. [10] who adjusted—similar to our protocol E—the tube current to the patients body constitution by means of attenuation-based tube current modulation.

Conclusion

Radiation dose of CS using dual-source CT ranges between 0.6 and 1.3 mSv, depending on the type of data acquisition and the protocol used. Prospective triggering significantly reduces the radiation dose delivered to patients as compared to the retrospective ECG-gating mode. A reduction of tube potential leads to significant reductions in radiation dose irrespective of whether the retrospective or prospective mode is applied. Both strategies may be implemented without an increase in image noise in normal weight patients. Further studies must aim at an assessment of the optimal CS protocol for dual-source CT with regard to the accuracy and reproducibility of coronary calcium burden quantification.

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