Original Research

Potentially Optimal Body Size to Adjust Tube Current for Individualized Radiation Dose Control in Retrospective ECG-Triggered 256-Slice CT Coronary Angiography

Jing-Lei Li[1,2], Hui Liu[2], Mei-Ping Huang[2], Zhong-Lin Zhang[2], Hong-Jun Liu[2], Yan-Hai Cui[2], Ling Xue[3], Chang-Hong Liang[1,2]

[1]Southern Medical University, Guangzhou, [2]Department of Radiology, [3]Department of Cardiology, Guangdong Academy of Medical Sciences, Guangdong General Hospital, Guangzhou, P.R. China

Key words: **Computed tomography coronary angiography, radiation dose, body size, chest circumference, tube current selection.**

Manuscript received: January 28, 2013; *Accepted:* November 26, 2013.

Address: Chang-Hong Liang

Department of Cardiology Guangdong Academy of Medical Sciences Guangdong General Hospital 106 Zhongshan Er Road Guangzhou P.R. China 510515
cjr.lchh@vip.163.com

Introduction: We aimed to determine a potentially optimal body size index for adjusting the tube current in retrospective ECG-triggered helical 256-slice CT coronary angiography (CTCA) for individualized radiation dose control.

Methods: Consecutive patients (n=102) with suspected coronary artery disease underwent retrospective ECG-triggered CTCA with a 256-slice CT scanner. Body mass index (BMI), nipple level (NL) bust and six anthropometric body size indexes, including thoracic anteroposterior diameter at NL, chest circumference (CC) at NL, left main (LM) and right coronary artery (RCA) origin level, chest area (CAr) and chest attenuation (CAt) at RCA origin level were measured, and their correlation with image noise in the aorta was evaluated using linear regression. Pearson correlation analysis was performed respectively to determine the body size index that correlated best with the other body size indexes. An equation was derived to use the best correlated body size index for adjusting tube current.

Results: Linear regression demonstrated that CCRCA had the best correlation with image noise. Pearson correlation analysis showed that CCNL, CCLM and CArRCA had the closest linear relationship with CCRCA. The equations connecting CCRCA and tube current for males and females were XmA = 662 × (0.055 × CCRCA - 28.594) / 302 and XmA = 662 × (0.049 × CCRCA - 21.584) / 302, respectively, for a fixed image noise value of 30 HU.

Conclusions: Tube current selection is different for males and females, particularly in patients with a small chest circumference. CCRCA is an ideal body index for appropriately adjusting tube current in CTCA for individualized radiation dose control.

With the more widespread use of multidetector computed tomography (MDCT) scanner technology as an important part of our everyday practice, this imaging technique has become the largest source of diagnostic medical radiation.[1] Since radiation exposure is known to be associated with a non-negligible lifetime attributable risk of breast or lung cancer,[2] strong concerns have arisen among physicians and radiologists regarding MDCT, and especially computed tomography coronary angiography (CTCA), which has emerged as a fast, accurate and noninvasive method for the evaluation of coronary artery disease (CAD).[3,4] Concerns about the high radiation dose in CTCA have hindered its widespread use.[2] Therefore, several strategies have been introduced to reduce the radiation dose during CTCA, including limiting Z axial scan length, minimizing the field of view (FOV), ECG tube current (mA) modulation, cardiac noise reduction filters, dynamic z-collimation, the development of high-pitch scanning, the use of prospective ECG-triggering and the use of iterative reconstruction.[5,6] In particular, the use of prospective ECG-triggering and the newer wide area detectors (e.g. 256-slice CT scanner) allow a substantial reduction of radiation exposure.[7,8]

However, there is considerable variability in the radiation dose in CTCA between individuals, because the degree of X-ray attenuation varies with different body size. Fox et al[9] reported that if the body size is reduced by 4 cm, the X-ray attenuation will be reduced by about 50% and image noise will be reduced by about 30%. Therefore, to optimize individualized radiation dose control, it is helpful to use an appropriate body size index for adjusting the radiation dose parameters, e.g. tube current and tube voltage. Presently, however, the radiation dose parameters are adjusted empirically, or based on body mass index (BMI) or weight,[10] which do not accurately account for individual differences in body size that are not often measured or recorded at the time of examination. It has been reported[11,12] that several anthropometric chest measurements might serve as a surrogate for BMI, but so far, the question as to which of them could serve as a marker of body size for adjusting dose settings is still controversial.[13]

The purpose of our study was to retrospectively determine which of the body size indexes (BMI or image-based anthropometric chest measurements) is most correlated with quantum mottle image noise in 256-slice CTCA and to create a linear equation between image noise and the best correlated body size index. By combining this with the equation connecting image noise and tube current, an equation linking the best correlated body size index and the tube current could be derived for individualized radiation dose control.

Methods

Study population

The study was approved by the local ethics committee at Guangdong General Hospital. Before the procedure, a cardiac radiologist (JL) explained the adverse effects of contrast medium injection and radiation exposure to all patients and obtained written informed consent.

The prospective study included 102 consecutive patients who underwent CTCA for suspected or known CAD between February 23rd and March 16th, 2010. Patients with previous bypass grafts, non-sinus rhythm, known premature ventricular or supraventricular beats, elevated serum creatinine levels (>150 μmol/L), or allergy to iodinated contrast agent were excluded. Nipple level bust (NLB) circumference (with tape), body weight, and height were measured manually by nursing staff in the radiology department. Patients without complete data (such as missing images, or weight or height data) were excluded from this study.

CTCA protocol

All examinations were performed on a wide detector 256-slice CT scanner (Brilliance iCT; Philips Healthcare, Cleveland, Ohio, USA) providing 8 cm of coverage. Patient preparation included administration of 25-50 mg oral metoprolol (AstraZeneca, Wuxi, China) 60 minutes before the CT scan, if a heart rate >75 beats/minute was measured before the scan. The heart rate was monitored throughout the whole CT scanning procedure and initial heart rate (IHR) was recorded. ECG-triggered calcium score scans were performed prospectively, and were reconstructed to measure chest circumference at the nipple level (CCNL), left main (CCLM) and right coronary artery origin level (CCRCA), the thoracic anteroposterior diameter at the nipple level (TAPDNL), and chest area and chest attenuation at the right coronary artery origin level (CArRCA, CAtRCA, respectively). All patients underwent retrospectively ECG-triggered scans, using the following protocol: tube voltage 120 kV, tube current 662 mA; collimation 128 × 0.625 mm; rotation time 270 ms; pitch 0.18; FOV 250 mm. The CT scanning range covered the whole heart from 1 cm below the tracheal bifurcation to 2 cm below the diaphragm. The standard temporal resolution of 135 ms was further improved by employing advanced cardiac adaptive multi-cycle reconstruction algorithms that combine data from consecutive cardiac cycles. The use of the overlapped pitch, along with dedicated cardiac gating algorithms (Beat-to-Beat Variable Delay Algorithm, Philips Healthcare, Cleveland, OH, USA)[14] enable the detection and reconstruction of the quiescent physiologic cardiac phase of interest.

Injection protocol

A volume of 80 mL of contrast medium (Ultravist 370, Bayer Schering Pharma, Berlin, Germany) followed by 30 mL saline solution was injected into an antecubital vein at a flow rate of 5 mL/s with an 18-gauge catheter using a double tube high-pressure syringe. A real-time bolus tracking technique (BolusPro, Philips Healthcare, Cleveland, Ohio, USA) was used, defining a region of interest (ROI) in the ascending aorta root, with the scans initiated 5 s after the signal attenuation reached a pre-determined threshold of 150 Hounsfield units (HU). Particular care was taken to exclude air bubbles from the connection tube.

Estimation of the radiation dose

The volume CT dose index (CTDIvol) in mGy and the dose-length product (DLP) in mGy.cm were obtained from the patient dose report. The effective dose (ED) in mSv was calculated as the product of the DLP multiplied by a conversion coefficient for the chest (k = 0.014 mSv / mGy.cm).[15]

Image reconstruction and analysis

Images were reconstructed using a standard reconstruction kernel (XCB) with a thickness of 0.9 mm, an increment of 0.45 mm, and a FOV of 250 mm, at 70%, 75%, and 80% of the RR interval if HR was <75 bpm, and at 40%, 45%, and 50% of the RR interval if HR was ≥75 bpm. All images were then transferred to an external workstation (Cardiac Viewer and Comprehensive Cardiac Analysis (CCA), Extended Brilliance Workspace (Version 4.0); Philips Healthcare, Cleveland, Ohio, USA) for interpretation. Images were evaluated in consensus by two trained cardiac radiologists (JL and MH), both with more than 5 years experience in CTCA. Three-dimensional techniques, such as curved planar reconstruction (CPR), multiplanar reformations (MPR), maximum intensity projection (MIP), and volume rendering (VR), were used for interpretation.

The image quality (IQ) of 14 coronary artery segments of ≥1.5 mm at their origin, including the left main coronary artery (LM), proximal, middle and distal left anterior descending coronary artery (LAD), first and second diagonal branch (D1, D2), proximal and distal left circumflex artery (LCX), left obtuse marginal branch (OM), proximal, middle and distal right coronary artery (RCA), posterior descending coronary artery (PDA), and posterior left ventricular branch (PL), were evaluated using a five-point scale (5: excellent image quality, i.e. no visible effects of noise and excellent contour and contrast; 4: good image quality, i.e. minimal visible effects of noise and good contour and contrast; 3: moderate image quality, i.e. mild visible effects of noise and moderate contour and contrast 2: poor image quality, i.e. moderate effects of noise and poor contour and contrast; 1: non-diagnostic image quality, i.e. high visible effects of noise and severe contour and contrast) (Figure 1). Scores 3–5 were considered as diagnostic, scores 1–2 as non-diagnostic.

To assess IQ in an objective fashion for each acquisition, the image noise was measured as the standard deviation of the pixels on the axial images in the middle of the aorta at the level of the LM origin in a 100 mm^2 circular ROI. In addition, the image noise, attenuation and contrast, as well as the signal-to noise ratio (SNR) and contrast-to noise ratio (CNR) of the proximal RCA and LM were determined for each dataset in a 2 mm^2 circular ROI. Background noise was defined as the standard deviation of CT density in the same ROI placed in the subcutaneous fat of anterior chest wall on the corresponding image. Calcified or other disturbing structures were carefully avoided.

Statistical analysis

Statistical analyses were performed using commercially available software (SPSS 13.0, USA). Continuous variables were expressed as mean ± standard deviation. A *t*-test for independent samples was used to respectively compare the body size indexes, age, IHR, image noise, SNR, CNR, and effective dose between males and females. A rank-sum test (Mann–Whitney test) was used to compare IQ score between males and females, with a level of $p<0.05$ considered as statistically significant. Linear regression was used to demonstrate the correlation between image noise in the aorta and all body size indexes, including BMI, NLB, CCNL, CCLM, CCRCA, TAPDNL, CArRCA, and CAtRCA, and to create a linear equation between the image noise in the aorta and the best correlated body size index. Pearson correlation analysis was also performed to respectively compare the best correlated body size index with the other body size indexes. Combined with the equation connecting image noise in the aorta and tube current, the equation between tube current and the best correlated body size index was derived, and then the tube

Figure 1. CT coronary angiography images of varying quality: excellent (1,2 – score 5), good (3,4 – score 4), moderate (5,6 – score 3), poor (7,8 – score 2), and non-diagnostic (9,10 – score 1).

current corresponding with the best correlated body size index was calculated for a fixed image noise level of 30 HU in the aorta.

Results

Data statistics and radiation exposure

In all 102 patients, CTCA was performed successfully without any adverse events, and all data were eligible for evaluation. All patients performed a successful breathhold during data acquisition. The detailed clinical and demographic data, body size indexes and radiation dose are shown in Table 1. No significant difference between males and females was observed in BMI, CCNL, CCLM, CCRCA, CArRCA, or CAtRCA ($p>0.05$), but NLB and TAPDNL were larger in males ($p<0.05$). The DLP and ED were significantly higher in males than in females ($p<0.05$).

Correlation between image noise in the aorta and indexes of body size

Among the males and females separately, as well as the entire cohort, linear regression analysis showed a strong correlation between image noise in the aorta and CCRCA, CCNL, CCLM, and CArRCA. The strongest correlation was with CCRCA (largest r^2), while weaker correlations were found with BMI, NBL and TAPDNL, and the weakest correlation was with CAtRCA. Correlation coefficients and corresponding p-values are shown in Table 2. Pearson correlation analysis showed that CCNL, CCLM and CArRCA had the closest linear relationship with CCRCA ($r>0.950$), BMI, NLB and TAPDNL had a weaker linear relationship, while there was no linear relationship between CAtRCA and CCRCA. The statistical results are shown in Table 3. The relationships between image noise in the aorta and CCRCA for males and females in our study were described by linear regression equations of the form:

$$y = 0.055x - 28.594 \ (r^2=0.605) \text{ and}$$

$$y = 0.049x - 21.584 \ (r^2=0.635),$$

respectively, where y represents the image noise in the aorta and x represents CCRCA. In our study, these equations were established while the tube current was fixed (FmA), so they could be respectively re-expressed as:

$$SD(FmA) = 0.055 \times CCRCA - 28.594, \text{ and}$$

$$SD(FmA) = 0.049 \times CCRCA - 21.584.$$

Table 1. Patient demographics and radiation dose estimate.

Variable	Value (mean ± SD and range)			t	p
	Male	Female	Male + Female		
Number	56	46	102		
Age (y)	60.0 ± 14.3 (17 to 88)	65.1 ± 12.8 (36 to 88)	62.3 ± 13.8	1.890	0.062
Weight (kg)	70.3 ± 11.6 (45 to 100)	59.1 ± 10.3 (35 to 81)	65.2 ± 12.3	5.053	0.000
Height (cm)	168.5 ± 5.7 (148 to 181)	157.5 ± 5.8 (145 to 78)	163.5 ± 8.0	9.676	0.000
BMI (kg/m^2)	24.6 ± 3.2 (18.0 to 32.2)	23.9 ± 4.0 (15.6 to 31.6)	24.3 ± 3.6	1.102	0.273
NLB (cm)	93.0 ± 6.9 (78 to 109)	89.8 ± 8.1 (75 to 115)	91.6 ± 7.6	2.214	0.029
CCLM (mm)	932.2 ± 64.7 (810.3 to 1079.6)	911.2 ± 81.3 (754.6 to 1165.6)	922.7 ± 73.0	1.450	0.150
CCRCA (mm)	925.4 ± 64.9 (804.8 to 1072.2)	907.0 ± 83.1 (750.1 to 1174.4)	917.1 ± 73.8	1.256	0.212
CCNL (mm)	927.6 ± 64.1 (803.6 to 1072.0)	913.8 ±80.8 (745.8 to 1171.1)	921.4 ± 72.1	0.961	0.339
TAPDNL (mm)	218.3 ± 18.9 (172.1 to 265.1)	207.2 ± 22.9 (150.7 to 252.2)	213.3 ± 21.4	2.684	0.009
CArRCA (mm^2)	614.4 ± 88.8 (452.0 to 831.8)	582.5 ± 98.2 (384.0 to 872.3)	600.0 ± 94.1	1.721	0.088
CAtRCA (HU)	-224.2 ± 66.9 (-376 to -56)	-210.3 ± 49.7 (-347 to -103)	-217.9 ± 59.9	1.169	0.245
IHR (bpm.)	71.0 ± 12.1 (45 to 98)	69.6 ± 10.9 (49 to 100)	70.3 ± 11.5	0.582	0.562
Noise (HU)	22.5 ± 4.6 (15.9 to 37.6)	22.8 ± 5.1 (13.6 to 39.6)	22.7 ± 4.8)	0.264	0.792
DLP (mGy.cm)	1202.7 ± 52.6 (1055.5 to 1435.9)	1162.3 ± 65.9 (1026.9 to 1209.9)	1184.5 ± 62.1	3.445	0.001
ED (mSv)	16.8 ±.07 (14.8 to 20.1)	16.3 ±.09 (14.4 to 16.9)	16.6 ± 0.9	3.445	0.001

BMI – body mass index; NLB – nipple level bust; CCLM – chest circumference at left main origin level; CCRCA – chest circumference at right coronary artery origin level; CCNL – chest circumference at nipple level; TAPDNL – thoracic anteroposterior diameter at nipple level; CArRCA – chest area at right coronary artery origin level; CAtRCA – chest attenuation at right coronary artery origin level; IHR – initial heart rate; DLP – dose-length product; ED – effective dose.

As we know, the equation between image noise in the aorta and tube current is expressed as $(SD)^2 = K / (V^3 \times H \times D)$ (SD: image noise in the aorta; V: voxel size; H: slice thickness; D: tube current; K: constant value). While V and H are constant, the equation between image noise in the aorta and fixed tube current (FmA) is re-expressed as $[SD(FmA) / INa(XmA)]^2 = (XmA / FmA)$ (INa: clinically selected image noise in the aorta; XmA: tube current corresponding with INa). Then, in our study, the equation connecting CCRCA and tube current could be derived and expressed as: $XmA = FmA \times [(K1 \times CCRCA + C1) / INa]^2$. While FmA was fixed at 662 mA and INa was fixed at 30 HU in our study, the equations between CCRCA and tube current for males and females were respectively reformulated as:

$$XmA = 662 \times [(0.055 \times CCRCA - 28.594) / 30]^2$$

and

$$XmA = 662 \times [(0.049 \times CCRCA - 21.584) / 30]^2$$

Then, on the basis of the above equations, the tube current corresponding to different values of CCRCA could be calculated (Table 4).

Subjective and objective image evaluation

A total of 1425 coronary artery segments were evaluated; three segments were missing because of anatomical variants. Among the 1425 coronary artery segments, IQ scores were ≥3 and were considered as diagnostic in 1423 segments (99.8%), whereas IQ scores were <3 and were considered as non-diagnostic in only 2 segments (0.2%). The corresponding IQ scores are shown in Table 5. Quantification analysis

Table 2. Statistical results of linear regression for eight body size indexes related to image noise.

Group	Body size index	r	r^2	t	p	constant	coefficient
Male	BMI	0.731	0.535	62.134	0.000	-3.734	1.067
	NLB	0.687	0.472	48.311	0.000	-20.361	0.461
	CCLM	0.769	0.591	77.932	0.000	-28.517	0.055
	CCRCA	0.778	0.605	82.744	0.000	-28.594	0.055
	CCNL	0.773	0.598	80.278	0.000	-29.036	0.056
	TAPDNL	0.711	0.505	55.060	0.000	-15.424	0.174
	CArRCA	0.770	0.593	78.773	0.000	-2.017	0.040
	CAtRCA	0.475	0.226	17.733	0.000	29.886	0.033
Female	BMI	0.737	0.543	52.248	0.000	0.442	0.937
	NLB	0.749	0.561	56.241	0.000	-19.708	0.474
	CCLM	0.794	0.630	75.057	0.000	-22.609	0.050
	CCRCA	0.797	0.635	76.518	0.000	-21.584	0.049
	CCNL	0.776	0.603	66.705	0.000	-21.988	0.049
	TAPDNL	0.639	0.408	30.321	0.000	-6.602	0.142
	CArRCA	0.773	0.597	65.273	0.000	-0.586	0.040
	CAtRCA	0.630	0.397	28.998	0.000	36.2594	0.065
Male + female	BMI	0.725	0.526	110.830	0.000	–1.073	0.977
	NLB	0.696	0.484	93.845	0.000	–17.832	0.442
	CCLM	0.768	0.590	144.158	0.000	–24.093	0.051
	CCRCA	0.776	0.603	151.649	0.000	–23.765	0.051
	CCNL	0.767	0.588	142.758	0.000	–24.547	0.051
	TAPDNL	0.642	0.412	70.116	0.000	–8.118	0.144
	CArRCA	0.756	0.567	133.095	0.000	-0,549	0.039
	CAtRCA	0.527	0.277	38.393	0.000	31.892	0.042

Abbreviations as in Table 1.

Table 3. Statistical results of Pearson correlation analysis for BMI, NLB, CCLM, CCNL, TAPDNL, CArRCA, and CAtRCA related to CCRCA.

	CCRCA					
	Male		Female		Male + female	
	r	p	r	p	r	p
BMI (kg/m^2)	0.869	0.000	0.884	0.000	0.879	0.000
NLB (cm)	0.805	0.000	0.878	0.000	0.845	0.000
CCLM (mm)	0.992	0.000	0.996	0.000	0.994	0.000
CCNL (mm)	0.972	0.000	0.987	0.000	0.980	0.000
TAPDNL (mm)	0.800	0.000	0.822	0.000	0.811	0.000
CArRCA (mm^2)	0.971	0.000	0.984	0.000	0.975	0.000
CAtRCA (HU)	0.348	0.009	0.488	0.001	0.375	0.000

Abbreviations as in Table 1.

including the SNR and CNR in the ascending aorta, proximal RCA, and LAD is shown in Table 6.

Discussion

Following the "as low as reasonably achievable" (ALARA) principle and maintaining sufficient image quality for a reliable CAD diagnosis, an individualized scanning protocol is an ideal, imperative method of performing CTCA, especially in retrospective ECG-triggered helical scans. Presently, BMI is the most commonly reported body index for adjusting tube voltage and tube current in CTCA, but it is not often measured or recorded at the time of examination and there is still debate whether it is the most appropriate and ideal body index for CTCA. In this regard, our study's aim was to compare different body size indexes for adjusting tube current selection in 256-slice MDCT with retrospective ECG-triggered helical CTCA, and to determine the most appropriate and ideal index for individualized radiation control while maintaining diagnostic image quality.

In our study, we found that CCRCA was the body

Table 4. The tube current corresponding to different chest circumference at right coronary artery origin level (CCRCA).

CCRCA	Tube current (mA)		CCRCA	Tube current (mA)	
	Male	Female		Male	Female
700	70	120	960	430	475
710	80	130	970	450	495
720	90	140	980	470	515
730	100	150	990	490	530
740	110	160	1000	510	550
750	120	170	1010	535	570
760	130	180	1020	555	590
770	140	190	1030	580	610
780	150	200	1040	600	630
790	160	215	1050	625	656
800	175	230	1060	650	680
810	190	240	1070	670	700
820	200	255	1080	700	720
830	210	265	1090	720	745
840	230	280	1100	760	770
850	240	295	1110	885	790
860	260	310	1120	800	815
870	270	325	1130	830	840
880	290	340	1140	855	864
890	305	355	1150	883	890
900	321	372	1160	910	915
910	340	290	1170	940	940
920	355	405	1180	970	965
930	375	420	1190	1000	990
940	390	440	1200	1030	1020
950	410	460	1210	1060	1045

Table 5. Image quality scores of each coronary artery segment in males and females.

	Male		Female		Z	p
	N	Image quality score	N	Image quality score		
RCA:						
Proximal	56	4.88 ± 0.38	46	4.98 ± 0.15	1.696	0.090
Mid	55	4.53 ± 0.60	46	4.65 ± 0.48	0.906	0.365
Distal	55	4.73 ± 0.53	45	4.69 ± 0.56	0.354	0.724
PD	56	4.86 ± 0.44	46	4.78 ± 0.51	0.946	0.344
PL	56	4.91 ± 0.39	46	4.85 ± 0.42	1.292	0.196
LM	56	4.96 ± 0.27	46	4.98 ± 0.15	0.126	0.900
LAD:						
Proximal	56	4.84 ± 0.46	46	4.98 ± 0.15	1.934	0.053
Mid	56	4.70 ± 0.57	46	4.89 ± 0.32	1.884	0.060
Distal	56	4.82 ± 0.47	46	4.78 ± 0.42	0.874	0.382
D1	56	4.88 ± 0.43	46	4.89 ± 0.32	0.261	0.794
D2	56	4.84 ± 0.50	46	4.89 ± 0.32	0.069	0.945
LCX:						
Proximal	56	4.88 ± 0.38	46	4.98 ± 0.15	1.696	0.090
Distal	56	4.82 ± 0.43	46	4.80 ± 0.40	0.412	0.680
OM	56	4.86 ± 0.40	46	4.78 ± 0.42	1.187	0.235

N – number of coronary artery segments; RCA – right coronary artery; PD – posterior descending coronary artery; PL – posterior left ventricular branch; LM – left main coronary artery; LAD – left anterior descending coronary artery; D1, D2 – first and second diagonal branch, LCX – left circumflex artery; OM – left obtuse marginal branch.

size index that best correlated with image noise in the aorta, among males, females, and the entire cohort, while CCNL, CCLM, and CArRCA were very close to or correlated with CCRCA. However, CAtRCA had only a weak correlation with image noise in the aorta. One major possible explanation is the large

Table 6. Objective image quality parameters in males, females, and the entire cohort.

Variable	Value (mean ± SD and range) Male	Female	Male + female	t	p
Mean HU in ascending aorta	391.0 ± 64.6 (253.0–544.0)	434.0 ± 80.9 (284.0–646.0)	410.6 ± 75.3 (253.0–646.0)	3.010	0.003
SNR in ascending aorta	18.6 ± 6.8 (1.6–38.6)	20.5 ± 6.5 (8.6–39.6)	19.5 ± 6.7 (1.6–39.6)	1.433	0.155
CNR in ascending aorta	33.1 ± 9.7 (17.6–62.1)	37.1 ± 12.5 (10.4–66.3)	34.9 ± 11.1 (10.4–66.3)	1.777	0.079
Mean HU in proximal RCA	330.6 ± 83.4 (140.0–479.0)	359.0 ± 79.8 (128.0–516.0)	343.0 ± 82.6 (128.0–516.0)	1.735	0.086
SNR in proximal RCA	13.7 ± 8.7 (4.6–56.5)	13.3 ± 9.2 (0.9–42.9)	13.5 ± 8.9 (0.9–56.5)	0.214	0.831
CNR in proximal RCA	26.4 ± 7.2 (14.6–46.3)	30.9 ± 10.3 (8.1–53.4)	28.5± 9.0 (8.2–53.4)	2.517	0.014
Mean HU in proximal LAD	316.1 ± 73.4 (177.0–477.0)	354.2 ± 112.2 (136.0–861.0)	333.0 ± 94.4 (136.0–861.0)	2.064	0.042
SNR in proximal LAD	14.1 ± 8.6 (3.9–44.2)	14.7 ± 11.0 (3.6–63.6)	14.4 ± 9.7 (3.6–63.6)	0.278	0.782
CNR in proximal LAD	28.2 ± 8.6 (13.1–51.8)	31.9 ± 12.6 (5.0–70.0)	29.9 ± 10.7 (5.0–70.0)	1.735	0.098

HU – Hounsfield units; SNR – signal-noise ratio; CNR – contrast-noise ratio; RCA – right coronary artery; LAD – left anterior descending coronary artery

proportion of low attenuation, air-filled lung tissue at the level of the heart, which could overwhelm the higher attenuation of the remainder of the chest.[13] The large manual measurement error with a tape to determine NLB might explain the relatively low correlation coefficient. The weaker correlation between TAPDNL and image noise in the aorta was mainly because TAPDNL was only one of the projections around the chest, whereas the thickness of the chest along a 0° to 360° projection varies greatly within each rotation during the helical scanning process for CTCA examinations. The fact that the correlation between BMI and image noise in the aorta was weaker than that of chest circumference and CArRCA may be due to differences in the composition of adipose versus muscular tissue or tissue out of the scan range, such as abdominal obesity and gluteal fat. In other words, BMI is known as a convenient proxy for human body fat, but it does not reliably represent human body shape, especially in patients with central obesity, whereas the scan range for CTCA is limited only to the chest or the heart. For instance, a large-chested woman might require more radiation than a small-chested man with the same BMI. Recently, Ghoshhajra et al[13] also reported that patients' chest area and BMI classes were frequently discordant, which could potentially lead to overdosing 21.4% of patients when using BMI to select tube potential. Women are known to have different quantities of adipose tissue than men.[16] From Table 4 in our study, we can see that the tube current in females tended to be higher than that in males even when they had the same CCRCA, with the difference being greater for smaller CCRCA values.

Our study suggested that there was a linear relationship between CCRCA and image noise in the aorta, i.e. the larger the CCRCA the higher the image noise. Therefore, to maintain high diagnostic IQ, patients with a larger CCRCA would require a higher tube current. On the other hand, the tube current in a CTCA examination differs between males and females, even those with the same CCRCA. A possible explanation is the larger proportion of subcutaneous fat (i.e. breast) and the relatively smaller proportion of air-filled lung tissue of the female compared to a male with the same chest circumference.

Our findings provide a simple, objective, standard, and precise method for using the patient's own images for patient-specific dose optimization and reduction, based on measurements taken from the calcium scoring scan for chest circumference, as opposed to relying on BMI measurements, which may not be representative of the area undergoing the CT scan.

Limitations

Our study has several limitations. Firstly, there is no randomized trial to further verify our conclusions; we plan to conduct such a trial protocol in the future. Secondly, we did not use prospective ECG-triggering, iterative reconstruction, tube current modulation and other dose reduction methods and techniques, but only retrospective ECG-triggering. However, in this study the purpose was to evaluate the ideal body size index for adjusting tube current selection to enable individualized radiation dose control; therefore, all the other methods and techniques had to be fixed. In future studies, we will use CCRCA to adjust the tube current, in combination with other dose reduction methods and techniques. Thirdly, the effective radiation was relatively higher than that previously reported.[17] The reason might be that we did not apply other dose reduction methods and techniques, so the tube current used in our study was relatively high. Finally, we did not analyze the diagnostic accuracy by using coronary angiography as a standard reference, because most patients did not undergo invasive coronary angiography; however, the diagnostic accuracy of the 256-slice scanner used in our study has been previously reported by Chao et al.[18]

Conclusions

Our findings suggest that anthropometric measures of chest circumference and chest area based on the patient's own images, especially CCRCA, are strongly correlated with image noise in the aorta. It may be better and more appropriate to use CCRCA in the adjustment of tube current in 256-slice MDCT retrospective ECG-triggered helical CTCA for individualized radiation dose control, as this could result in a lower radiation dose, especially in smaller patients. In addition, the tube current selection is different in males and females, particularly in those patients with a small chest circumference. Larger and prospective randomized trials are needed to further verify our results.

Acknowledgements

This work was partly supported by Guangdong Province Medical Research Foundation (A2013036), Science and Technology Planning Project of Guangdong Province (2009B030801257, 2006B36030026), Science and Technology Planning Project of Guangzhou City (2006Z3-E241).

References

1. Stefanadis CI. Ionising radiation: not the big bad wolf, but definitely not little red riding hood. Hellenic J Cardiol. 2012; 53: 405-406.
2. Einstein AJ, Henzlova MJ, Rajagopalan S. Estimating risk of cancer associated with radiation exposure from 64-slice computed tomography coronary angiography. JAMA. 2007; 298: 317-323.
3. Rybicki FJ, Otero HJ, Steigner ML, et al. Initial evaluation of coronary images from 320-detector row computed tomography. Int J Cardiovasc Imaging. 2008; 24: 535-546.
4. Motoyama S, Anno H, Sarai M, et al. Noninvasive coronary angiography with a prototype 256-row area detector computed tomography system: comparison with conventional invasive coronary angiography. J Am Coll Cardiol. 2008; 51: 773-775.
5. Gopal A, Budoff MJ. A new method to reduce radiation exposure during multi-row detector cardiac computed tomographic angiography. Int J Cardiol. 2009; 132: 435-436.
6. Ghadri JR, Küest SM, Goetti R, et al. Image quality and radiation dose comparison of prospectively triggered low-dose CCTA: 128-slice dual-source high-pitch spiral versus 64-slice single-source sequential acquisition. Int J Cardiovasc Imaging. 2012; 28: 1217-1225.
7. Herzog BA, Husmann L, Valenta I, et al. Determinants of vessel contrast in BMI-adapted low dose CT coronary angiography with prospective ECG-triggering. Int J Cardiovasc Imaging. 2009; 25: 625-630.
8. Huang MP, Liang CH, Zhao ZJ, et al. Evaluation of image quality and radiation dose at prospective ECG-triggered axial 256-slice multi-detector CT in infants with congenital heart disease. Pediatr Radiol. 2011; 41: 858-866.
9. Fox SH, Toth T. Dose reduction on GE CT scanners. Pediatr Radiol. 2002; 32: 718-723.
10. Hosch W, Stiller W, Mueller D, et al. Reduction of radiation exposure and improvement of image quality with BMI-adapted prospective cardiac computed tomography and iterative reconstruction. Eur J Radiol. 2012; 81: 3568-3576.
11. Menke J. Comparison of different body size parameters for individual dose adaptation in body CT of adults. Radiology. 2005; 236: 565-571.
12. Kalra MK, Maher MM, Prasad SR, et al. Correlation of patient weight and cross-sectional dimensions with subjective image quality at standard dose abdominal CT. Korean J Radiol. 2003; 4: 234-238.
13. Ghoshhajra BB, Engel LC, Major GP, et al. Direct chest area measurement: A potential anthropometric replacement for BMI to inform cardiac CT dose parameters? J Cardiovasc Comput Tomogr. 2011; 5: 240-246.
14. Vembar M, Garcia MJ, Heuscher DJ, et al. A dynamic approach to identifying desired physiological phases for cardiac imaging using multislice spiral CT. Med Phys. 2003; 30: 1683-1693.
15. Bischoff B, Hein F, Meyer T, et al. Impact of a reduced tube voltage on CT angiography and radiation dose: results of the PROTECTION I study. JACC Cardiovasc Imaging. 2009; 2: 940-946.
16. Freedman DS, Jacobsen SJ, Barboriak JJ, et al. Body fat distribution and male/female differences in lipids and lipoproteins. Circulation. 1990; 81: 1498-1506.
17. Hirai N, Horiguchi J, Fujioka C, et al. Prospective versus retrospective ECG-gated 64-detector coronary CT angiography: assessment of image quality, stenosis, and radiation dose. Radiology. 2008; 248: 424-430.
18. Chao SP, Law WY, Kuo CJ, et al. The diagnostic accuracy of 256-row computed tomographic angiography compared with invasive coronary angiography in patients with suspected coronary artery disease. Eur Heart J. 2010; 31: 1916-1923.