

# Automatic tube current modulation technique for multidetector CT: is it effective with a 64-detector CT?

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**Abstract** To investigate whether it is possible to obtain adequate images at uniform image noise levels and reduced radiation exposure with our automatic tube current modulation (ATCM) technique for 64-detector CT. The study population consisted of 64 patients with known or suspected lung or abdominal disease. We used a 64-detector CT scanner (LightSpeed VCT, GE Healthcare, Waukesha, WI, USA) and a combined angular and longitudinal tube current modulation technique (Smart mA, GE Healthcare, Waukesha, WI, USA) to examine 34 patients. The scanning parameters were identical; the minimum and maximum tube current thresholds were 50 and 800 mA, respectively. For study of the constant tube current technique, 30 additional patients were examined at 350 mA. The CT number and image noise (SD of the CT number) were measured in the 64 patients at six levels, i.e., the center of the left ventricle, the liver dome, the porta hepatis, the center of the spleen and the right and left renal pelvis. When we used the ATCM technique, the mean image noise ranged from 8.40 at the center of the left ventricle to 11.31 at the porta hepatis; the mean tube current ranged from 105.9 mAs at

the center of the left ventricle to 169.6 mAs at the center of the spleen. The mean dose reduction rate per constant tube current at 175 mAs ranged from 3.1 to 39.5%. By use of the ATCM technique, it is possible to maintain a constant image noise level with a 64-detector CT.

**Keywords** Automatic tube current modulation (ATCM) technique · Noise index · Radiation dose · Computed tomography (CT)

## 1 Introduction

Automatic tube current modulation (ATCM) with multi-detector CT (MDCT) is analogous to the automatic exposure control and photo-timing techniques that have been used for automatically terminating radiographic exposure in conventional radiography [1–5]. This modulation technique adjusts the tube current from slice to slice depending on the regional body anatomy, thereby maintaining a constant user-specified quantum image noise level and improving the radiation-dose efficiency. It involves angular or  $x$ - $y$  modulation,  $z$ -axis modulation, and combined or  $x$ - $y$ - $z$ -axis modulation. The combined angular and  $z$ -axis modulation technique has been introduced on commercially available MDCT scanners and represents the most comprehensive approach to CT dose reduction because the X-ray dose is adjusted to reflect patient-specific attenuation in all three planes [6, 7].

Because of technical improvements, the detector width along the  $z$  axis of MDCT scanners continues to increase. The detector width of our first 4-detector CT was 20 mm; the width of the 64-detector CT currently in use at our institute is 40 mm. Furthermore, as the tube rotation time of 0.8 s in 4-detector CT has been reduced to 0.5 s for

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64-detector instruments, the volume coverage speed is now as much as 12.8 times faster. On these newer instruments featuring a greater volume coverage speed, automatic dose modulation via angular modulation may not work well in areas with different adjacent attenuation values, for example, at the interface of the lungs and abdomen, because the tube current is determined according to projection data 180° from the X-ray generation angle. Our purpose of this study was to investigate whether the ATCM technique, applied to 64-detector CT, can effectively provide images with uniform image noise at reduced radiation doses.

## 2 Materials and methods

### 2.1 Patients

The human research committee of our institutional review board approved our study; prior informed consent was obtained from all patients. Between April 2007 and May 2007, we enrolled 30 patients (15 women and 15 men, mean age  $60.6 \pm 15.8$  years, range 23–85 years) in studies to assess the constant tube current (CTC) technique. Between June and July 2007, we enrolled another 34 patients (15 women and 19 men, mean age  $60.4 \pm 18.3$  years, range 20–84 years) in studies to evaluate the ATCM technique. All patients had known or suspected lung or abdominal disease; all had undergone plain chest and abdominal CT studies. We studied patients with available plain CT scans because image noise (standard deviation of the CT number) for enhancement CT was affected corresponding to the dose rate and the total dose of contrast medium.

### 2.2 Automatic tube current modulation technique

We used a 64-detector CT scanner (LightSpeed VCT, GE Healthcare, Waukesha, WI, USA). The scanning parameters were: tube voltage, 120 kVp; detector collimation,  $64 \times 0.625$  mm; beam pitch, 0.984; table speed, 39.37 mm per gantry rotation (gantry rotation time, 0.5 s); acquisition of 5.0 mm images reconstructed at 5.0 mm intervals; scan field of view (FOV), 50.0 cm; display FOV, 36.0 cm; matrix,  $512 \times 512$ ; standard reconstruction kernel. In our assessment of the combined angular and longitudinal tube current modulation technique (Smart mA, GE Healthcare, Waukesha, WI, USA), we applied identical scanning parameters to study 34 patients; the minimum and maximum tube current thresholds were 50 (25 mAs) and 800 mA (400 mAs), respectively. The noise index was set at 10 HU according to our routine protocol [8–10].

The mean body height and weight of the 34 patients were  $161.4 \pm 10.2$  cm and  $58.94 \pm 13.53$  kg, respectively.

### 2.3 Constant tube current technique

We used the same 64-detector CT scanner to assess the CTC technique in 30 patients. The scans were acquired at 350 mA, 120 kVp (175 mAs); the CTDIvol was 15.21 mGy [11, 12]. Other parameters were the same as in our study of the ATCM technique [6, 7].

### 2.4 Quantitative data

For each patient, we recorded the maximum transverse (mean 30.0 cm at ATCM and 29.1 cm at CTC) and anteroposterior (mean 22.0 cm at ATCM and 21.5 cm at CTC) diameter of the abdomen on transverse images obtained at the liver dome, using a DICOM viewer (Image Vinus Pro, ver. 3.01, Yokogawa, Tokyo, Japan) and a color monitor (Radiforce R22, Eizo, Ishikawa, Japan). The CT number and image noise, represented by the standard deviation of the CT number, were measured in all 64 patients (constant mA, 30 patients; variable mA, 34 patients) at six levels, i.e., the center of the left ventricle, the liver dome, the porta hepatis, the center of the spleen and the right and left renal pelvis, by use of the same DICOM viewer and color monitor. A total of 396 images (66 patients  $\times$  6 levels) were measured by use of the region of interest and contrast shape (circular) and size (40 square pixels). Each image was measured at least three times for calculating the mean value. In our study of the ATCM technique, we recorded the tube current at each level of the image; the body mass index (BMI) was calculated with the formula  $\text{BMI} = \text{weight (kg)} / \text{height (m)}^2$  divided by the square of the height (m) [13]. We categorized the BMI as follows: category 1:  $\text{BMI} \leq 18.5 \text{ kg/m}^2$  (underweight); category 2: BMI between 18.5 and 24.9  $\text{kg/m}^2$  (normal weight); category 3: BMI between 25 and 29.9  $\text{kg/m}^2$  (overweight); category 4:  $\text{BMI} \geq 30 \text{ kg/m}^2$  (obese). BMI is an index of human constitution based on height and weight and a worldwide method of categorizing human constitution with the BMI is established [14, 15]. Therefore, we used the BMI for a parameter, which evaluates human constitution.

### 2.5 Statistical analysis

The image noise was calculated, and values obtained in the center of the left ventricle, the liver dome, the porta hepatis, the center of the spleen and the right and left renal pelvis with and without the ATCM technique were

compared using Student's *t* test. The image noise at different BMI values was compared by one-way analysis of variance (ANOVA). When the overall differences were statistically significant, we performed a post-hoc analysis using Dunnett's method; the image noise at a BMI of 18.5–24.9 kg/m<sup>2</sup> (normal weight) was used as control. Differences at  $P < 0.05$  were considered statistically significant. To apply Student's *t* test, we used one-way ANOVA and the post-hoc test and a statistical software program (SPSS, version 10.05, SPSS, Chicago, IL, USA).

### 3 Results

Table 1 compares the mean CT number, the mean image noise (SD of the CT number), and the mean SD of the image noise obtained at the six examined levels with the ATCM and the CTC technique. At 6.33 HU, the mean image noise with the CTC was lowest at the center of the left ventricle; it was 7.40 HU at the liver dome and 10.16 HU at the porta hepatis on images obtained with the CTC technique at 175 mAs. At the six levels examined, the CTC technique consistently yielded significantly lower image noise values than did the ATCM technique ( $P < 0.05$  or  $P < 0.001$ ).

With the ATCM technique, the mean image noise ranged from 8.40 HU at the center of the left ventricle to 11.31 at the porta hepatis. Although the mean image noise was lower at the center of the left ventricle and the liver dome than at the other levels examined, these values were close to the 10 HU noise index. At 175 mAs, the mean SD of the image noise was lower with the ATCM than with the CTC technique (1.09–1.39 HU vs. 1.20–2.56 HU).

Table 2 shows the tube current reduction rate obtained with the ATCM technique. The mean tube current ranged from 105.9 mAs at the center of the left ventricle to 169.6 mAs at the center of the spleen. The mean reduction rate per constant tube current at 175 mAs ranged from

3.1% (at the center of the spleen) to 39.5% (at the center of the left ventricle). The difference in the range of the tube current was greatest in the right and left renal pelvis [range, 36.0–325.5 mAs; difference, 289.5 mAs vs. 288.0 mAs (the center of the left ventricle, the porta hepatis, the center of the spleen) and 284.0 mAs (liver dome)].

The mean tube current at the four different BMI classifications is shown in Table 3. At the center of the spleen, the difference between BMI category 1 (underweight) and category 4 (obese) with respect to the mean tube current was 236.5 mAs (range, 78.3–314.8 mAs); it was 223.0 mAs for the center of the left ventricle (range, 39.8 vs. 262.8 mAs). As shown in Table 4, the difference between BMI category 1 and category 4 with respect to image noise at the porta hepatis was 0.91 HU (range, 10.68–11.59 HU); it was 4.52 HU (range, 7.29–11.81 HU) at the liver dome. When we evaluated the four BMI categories with one-way ANOVA, the overall difference in the mean image noise at each of the six levels examined was statistically significant ( $P < 0.001$ ). In the liver dome, the mean image noise in BMI category 1 (7.29 HU) was significantly lower than in category 2 (8.87 HU;  $P < 0.001$ ). At the center of the left ventricle, the liver dome and the right and left renal pelvis, the mean image noise in obese patients (BMI category 4) was significantly higher than those in patients of normal weight (BMI category 2;  $P < 0.001$ ).

### 4 Discussion

In our CT scanner, the algorithm for ATCM is designed to maintain a constant image noise level as the attenuation values change from one rotation to the next. Our initial hypothesis posited that whereas the ATCM technique would fail to control the tube current at the interface of the lung and abdomen, i.e., at the liver dome, at the six levels studied, the tube current could be controlled and an almost

**Table 1** Comparison of the mean CT number, the mean image noise and the mean SD of the image noise with ATCM and CTC techniques

	ATCM technique			CTC technique		
	CT number (HU)	Image noise (HU)	SD (HU) of the image noise	CT number (HU)	Image noise (HU)	SD (HU) of the image noise
The center of the left ventricle	40.21	8.40	1.09	38.83	6.33**	1.20
The liver dome	54.68	8.59	1.39	56.16	7.40**	1.88
The porta hepatis	57.71	11.31	1.12	58.98	10.16*	2.55
The center of the spleen	48.49	11.11	1.13	47.15	9.70**	2.35
The right renal pelvis	34.55	10.74	1.11	35.16	9.56**	2.56
The left renal pelvis	35.44	10.69	1.30	34.81	9.54**	2.55

Mean image noise is defined as average standard deviation (SD) of the CT number

\*  $P$  value  $\leq 0.05$ , \*\*  $P$  value  $\leq 0.001$ ; Student's *t* test

**Table 2** Tube current reduction rate obtained with automatic tube current modulation (ATCM) technique

	Tube current in ATCM technique (mAs)		Dose reduction rate (%) <sup>a</sup>	
	Mean	Range	Mean	Range
The center of the left ventricle	105.9	25.5–313.5	39.5	85.7–179.1
The liver dome	144.0	33.0–317.0	17.7	81.1–181.1
The porta hepatis	167.4	36.0–324.0	4.3	79.4–185.1
The center of the spleen	169.6	36.0–324.0	3.1	79.4–185.1
The right renal pelvis	167.4	36.0–325.5	4.3	79.4–186.0
The left renal pelvis	160.9	36.0–325.5	8	79.4–186.0

<sup>a</sup> Dose reduction rate when 175 mAs with CTC technique was used as a standard

**Table 3** Mean tube current (mAs) at classified body mass indexes (BMI)

	Mean tube current (mAs)			
	BMI category 1 (underweight) <18.5 kg/m <sup>2</sup> (n = 5)	BMI category 2 (normal weight) 18.5–24.9 kg/m <sup>2</sup> (n = 18)	BMI category 3 (overweight) 25.0–29.9 kg/m <sup>2</sup> (n = 7)	BMI category 4 (obese) >30 kg/m <sup>2</sup> (n = 4)
The center of the left ventricle	39.8	91.1	151.9	262.8
The liver dome	69.3	124.7	197.1	313.3
The porta hepatis	78.6	149.7	232.9	317.5
The center of the spleen	78.3	152.0	233.7	314.8
The right renal pelvis	77.5	147.3	239.7	320.0
The left renal pelvis	74.2	143.5	230.1	320.8

**Table 4** Mean image noise for the four body mass index (BMI) categories

	Mean image noise (HU)			
	BMI category 1 (underweight) <18.5 kg/m <sup>2</sup> (n = 5)	BMI category 2 (normal weight) 18.5–24.9 kg/m <sup>2</sup> (n = 18)	BMI category 3 (overweight) 25.0–29.9 kg/m <sup>2</sup> (n = 7)	BMI category 4 (obese) >30 kg/m <sup>2</sup> (n = 4)
The center of the left ventricle	8.45 (0.68)	8.22 (1.08)	8.19 (0.78)	10.75 (1.12)**
The liver dome	7.29 (0.72)**	8.87 (1.10)	8.07 (0.50)	11.81 (2.02)**
The porta hepatis	10.68 (0.58)	11.36 (1.28)	11.53 (0.98)	11.59 (1.35)
The center of the spleen	10.70 (1.18)	10.95 (1.05)	10.96 (0.86)	12.09 (1.19)
The right renal pelvis	10.83 (1.52)	10.50 (0.94)	10.77 (0.90)	12.60 (0.21)**
The left renal pelvis	9.58 (1.21)	10.30 (1.06)	11.94 (0.88)	12.23 (0.71)**

Data are represented as image noise and standard deviation (SD) of the image noise

\*  $P$  value  $\leq 0.05$ , \*\*  $P$  value  $\leq 0.001$ ; Student's  $t$  test

constant image noise level could be maintained, even at the lung–abdomen interface. In the current study, we set the helical pitch at 0.984 because, at helical pitch values below 1.0, redundant projection data can be acquired and aliasing artifacts can be prevented [16]. We think that small pitch values less than 1.0 are effective for the ATCM to function well in even the portion where materials with very different attenuations are adjacent. On the contrary, the ATCM technique may fail to control the tube current if a pitch value larger than 1.0 is adopted. We are now investigating

what is the maximum pitch for the effective use of the ATCM technique.

A prototype of a 256-detector CT scanner features a detector width of 128 mm [17]. At that width, techniques other than the current ATCM method must be devised to allow for optimization of the radiation dose.

In general, the image noise is inversely proportional and the radiation dose is proportional to the tube current [18–20]. With the ATCM technique, when the maximum tube current is sufficiently high, the image noise remains

constant regardless of the patient BMI or weight. On the other hand, when the value of the maximum tube current is low, the image noise in overweight and obese patients may increase, preventing the acquisition of diagnostic images. We set the maximum tube current at 800 mA (400 mAs) in our assessment of the ATCM technique. We found that the image noise in overweight and obese patients was significantly higher than that in normal weight and underweight patients, although with the ATCM technique it was possible to maintain an almost constant image noise level in these individuals. Thus, when the ATCM technique is used, the maximum tube current should be based on both the image noise (image quality) and the radiation dose. Studies are underway in our laboratory for identifying the optimal maximum tube current.

In further attempts at radiation-dose reduction without degradation of the image quality, we must take into account the body and lesion size and contrast, the section thickness and the window level and width on the display monitor. However, these factors are not considered in the ATCM technique because it is difficult to assess their effects on lesion detection. Because dose modulation based on the contrast–noise ratio (CNR) may allow a further reduction in the radiation dose, we are investigating the relationship between lesion detectability and the CNR at various scan settings.

Our results are encumbered by some limitations. First, in our CTC scans we used 350 mA (175 mAs), which, in our system, is routine for abdominal scanning. Although the use of different tube currents may alter the dose reduction rate, we posit that 175 mAs represents an acceptable value because the mean tube current with the ATCM technique was 152.5 mAs. Second, the distribution of individuals ( $n = 34$ ) among the four BMI categories was not uniform because the collection of patients with evenly distributed body types is difficult. To acquire a large study population, we are in the process of designing a multicenter study.

In conclusion, with the ATCM technique it is possible to maintain a constant image noise level with 64-detector CT regardless of the patient BMI. For optimizing the radiation dose for the next generation of CT scanners, such as the 256-detector instrument, new methods need to be developed.

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