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Automatic exposure control at MDCT based on the contrast-to-noise ratio: Theoretical background and phantom study

Yoshinori Funama^{a,*}, Yoshiaki Sugaya^b, Osamu Miyazaki^b,
Daisuke Utsunomiya^c, Yasuyuki Yamashita^c, Kazuo Awai^d

^a Department of Medical Physics, Faculty of Life Sciences, Kumamoto University, Kumamoto, Japan

^b Design Department, CT Systems Division, Hitachi Medical Corporation, Kashiwa, Japan

^c Department of Diagnostic Radiology, Graduate School of Medical Sciences, Kumamoto University, Kumamoto, Japan

^d Department of Diagnostic Radiology, School of Biomedical Sciences, Hiroshima University, Hiroshima, Japan

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KEYWORDS

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Abstract *Purpose:* To develop a new automatic exposure control (AEC) technique based on the contrast-to-noise ratio (CNR) and provide constant lesion detectability.

Methods: Lesion detectability is affected by factors such as image noise, lesion contrast, and lesion size. We performed ROC analysis to assess the relationship between the optimum CNR and the lesion diameter at various levels of lesion contrast. We then developed a CNR-based AEC algorithm based on lesion detectability. Using CNR-based AEC algorithm, we performed visual evaluation of low-contrast detectability by 5 radiologists on a low-contrast module of the Catphan phantom, a contrast-difference level of 1.0% (difference in the CT number = 10 HU), and objects 3.0–9.0 mm in diameter.

Results: On step-and-shoot scans the mean detection fraction with CNR-based AEC remained almost constant from 88 to 99 % regardless of the lesion size. We observed the same trend on helical scans, the mean detection fraction with CNR-based AEC exhibited a high score from 91 to 100%. Although CNR-based AEC maintains higher CNR for smaller size or lower contrast lesion, radiation dose on 3 mm lesion resulted in about 13 times larger than that of 9 mm lesion size. CTDI_{vol} for the CNR-based AEC technique changed dramatically with the SD_z from 7.5 to 100.0 mGy for step-and-shoot scans and from 9.1 to 121.5 mGy for helical scans.

Conclusions: From the viewpoint of ROC analysis-based CNR for lesion detection, CNR-based AEC potentially provide image quality advantages for clinical implementation.

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* Corresponding author. Tel./fax: +81 96 373 5455.

E-mail address: funama@kumamoto-u.ac.jp (Y. Funama).

Introduction

The current automatic exposure control (AEC) technique at multi-detector computed tomography (MDCT) regulates the radiation dose to improve the consistency of image quality among patients and to provide better control of the absorbed dose [1–7]. Algorithms used by such AEC techniques (SD-based AEC) are designed to yield an operator-desired image noise (standard deviation of the CT number). In daily practice, operator will input the acceptable image noise level compatible with diagnostic purposes to detect the expected size and contrast of the lesions. Although lesion detectability at MDCT is affected not only by the image noise level but also the lesion-to-background contrast (C_{LB}) [8–10], current AEC techniques do not take C_{LB} into consideration. Moreover, there is the poor relevance of image noise with respect to image quality, other quality indices should be taken into account. In general, C_{LB} is related to the contrast-to-noise ratio (CNR), and CNR has been considered to be more closely related to image quality than image noise. Kalender et al. [11] examined the dose-weighted CNR to determine the optimal energy for coronary CT angiography and cardiac- and pediatric CT. Their results indicate that the optimal tube voltage has to be changed based on CNR to minimize the delivered dose. Thus, CNR is an important index for optimization parameters with respect to image quality and patient dose exposure.

Detectability of low-contrast tissues is crucial in routine CT especially when looking for small lesions in abdominal organs such as liver, spleen, pancreas or kidneys [12]. Therefore AEC based on the CNR that considers C_{LB} may be a better method for regulating the radiation dose for lesion detection. However, CNR-based AEC is not available because the desired CNR level for lesion detection [13] remains not to be determined quantitatively. Therefore, we investigated the optimum CNR required for the detection of lesions of different sizes and different C_{LB} values in efforts to develop a new AEC algorithm based on CNR.

Materials and methods

Outline of the CNR-based AEC algorithm

A flowchart to calculate the desired standard deviation [SD] of CT values (i.e. the image noise) along a patient z-axis using the CNR-based AEC algorithm is shown in Fig. 1. First, a scan projection radiograph (scanogram) is obtained to assess the patient attenuation features along the z-axis; this information can be translated into a water equivalent thickness at each z-axis position. Then the lesion-to-background contrast at each z-axis position ($C_{LB,Z}$) is calculated by entering the predefined C_{LB} value (C_{PRE}) into a $C_{LB,Z}$ conversion table. C_{PRE} indicates an assumed C_{LB} for lesion detection under standard conditions such as routine tube voltage and typical patient size at contrast-enhanced CT examinations. Here, $C_{LB,Z}$ was calculated with the equation:

$$C_{LB,Z} = C_{COEF} * C_{PRE}, \quad (1)$$

Figure 1 Flowchart showing the process of calculating the desired image noise along a z-axis using the CNR-based AEC algorithm.

where C_{COEF} is the conversion coefficient determined as in a previous study [14] (see Appendix). This equation indicates that $C_{LB,Z}$ varies as a function of the tube voltage and patient size.

Based on the lesion diameter and $C_{LB,Z}$ derived with equation (1), the detectable CNR (dCNR) at each z-axis position ($dCNR_z$) was calculated. Here, C_{PRE} and the lesion diameter are determined by the user and based on the desired lesion condition for detection. The lesion diameter must be large enough for the diagnostic task. Consequently, dCNR represents the specific CNR values considered acceptable for detecting lesions of a given diameter and $C_{LB,Z}$. Based on the results of the observer performance test, dCNRs were obtained under different visual conditions for 4 lesion diameters and 3 C_{LB} levels. Note that CNR is defined as the contrast divided by the image noise. Now values for $C_{LB,Z}$ and $dCNR_z$ can be obtained and the image noise at each z-axis position (SD_z) is calculated with the equation:

$$SD_z = \frac{C_{LB,Z}}{dCNR_z}, \quad (2)$$

Finally, combined angular and longitudinal tube current modulation is applied to achieve the desired SD_z .

Determination of the optimal CNR

Simulation of the CT scan

We used a mathematically-computed simulated phantom [15] in Fig. 2. It consists of cylindrical water containing a cylindrical lesion whose attenuation was lower than the water region. The parameters for the virtual CT scan were: rotation time 0.8 s, beam width at the center of rotation 0.625 mm, image thickness 0.625 mm, display field of view (FOV) 350 mm, matrix 512×512 . Step-and-shoot scans (non-helical) were acquired and CT images were

Figure 2 Example of the simulated CT images generated in this experiment. The lesion is located at the arrowhead position. It is a mathematically-designed cylindrical object that produced no partial volume effects. Within a 160-mm radius of the dashed circle, the image noise at any position is almost the same. We confirmed that variations in image noise were generally less than $\pm 2\%$ of the desired value.

generated by an abdominal standard reconstruction kernel (F32) available from ECLOS (Hitachi Medical Corp., Chiba, Japan). The average CT number of the water region was set to just about zero; the noise level was controlled by changing the virtual tube current. Individual images did or did not contain a single lesion, representing a nodule, at random locations within the 320-mm diameter dashed circle. We confirmed that the errors between the desired noise level and the measured image noise of water region at arbitrary positions inside the circle were less than 2%. The other simulation parameters are described in Table 1. The detectability of a lesion was manipulated by changing the simulation conditions on 2240 images. The diameter of the lesions varied from 3.5-, 5-, 10-, to 15 mm. CNR and C_{LB} also varied in lesions of the same size and CNR is expressed as follows: $CNR = C_{LB}/SD$. The CNR in 3.5-, 5-, 10-, and 15-mm lesions ranged from 3.00 to 4.50, 1.50–3.00, 0.50–1.50, and 0.40–1.10, respectively. C_{LB} values were set to 15–20, 5–20, 5–20, and 5–20 HU for 3.5-, 5-, 10-, and 15-mm lesions, respectively. In small lesions measuring 3.5 mm in diameter, C_{LB} at 5–10 yielded very low contrast. We consider these conditions less meaningful in the clinical situation and omitted the simulation parameters. For 3.5-mm diameter lesions, 16 sets of 20 images were generated; for 5–15 mm diameter lesions we generated 96 sets of 20 images. Each set consisted of 10 images without and with a single lesion at random locations.

Observer performance test

For the observer performance test we prepared a total of 2240 CT images. Five board-certified radiologists with

6–22 years of experience independently assessed the images in a blinded manner. The time for reading the images was not limited. The applied window width of 250 HU was based on routine abdominal examination protocols and the applied window level of 0 HU was adjusted to the level of the water CT value. The images were read in 4 sessions at which the observers were presented with lesions of different diameters. In each session they read 320–640 images. To compensate for reading bias during the same session, the images were presented in random order. Before each session, the radiologists were shown 64 images of different CNR and C_{LB} simulation parameters for training to ensure consistency of observer's criteria during the same session.

A continuous rating scale consisting of a line-marking method was used. This method requires observer to express their opinion in a visual form whether an object was present or not. In practice, each observer placed a digital mark at an appropriate position on a digital track bar on the custom software in each image. The software recorded a judgment of "definitely present" or "definitely not present" by means of reading the digital marked position at the right or left end of the track bar, respectively.

A binormal receiver operating characteristic (ROC) curve was fitted to each radiologist's confidence rating data with quasi-maximum likelihood estimation. A computer program (ROCKIT, version 0.9.1; University of Chicago, Chicago, IL) was used to obtain binormal ROC curves from the ordinal-scale rating data.

Determination of detectable CNR

We chose detectable CNR (dCNR) as the criterion for the CNR-based AEC algorithm. An approach for determining the dCNR for each lesion diameter is shown below. The procedure for generating the ROC curves derived from the observers' confidence rating results. For each observer there were 80 data sets for ROC analysis; they consisted of 8 data sets for 3.5- and 72 data sets for 5–15 mm diameter lesions. To obtain a sufficient number of samples, each data set included 20 images with- and 20 without a lesion; this was achieved by integrating two C_{LB} conditions (5–10-, 10–15-, and 15–20 HU) except for the 3.5 mm diameter lesion. As shown in Table 1, we only combined the data of 15 and 20 HU for the 3.5 mm diameter lesion.

ROC curve shows the functional relationship between the conditional probabilities of the presence and absence of a lesion (true-positive fraction: TPF, false-positive fraction: FPF). Since ROC analysis was done at each CNR and integrated 2 C_{LB} conditions, 8 and 72 curves were obtained for lesions with a 3.5- and 5–15 mm diameter, respectively. These 80 ROC curves were obtained by averaging the 5 observers' ROC curves. In each ROC curve reflecting 8 CNR conditions, the FPF corresponding to the specific TPF value is identified. In this evaluation, we chose a threshold that corresponded to a TPF of 0.95. Then we determined the relationship between CNR and FPF. On the basis of this relation, we calculated the CNR corresponding to the specific value (0.05) of FPF. In this way we determined dCNR for each lesion diameter and each C_{LB} condition.

Table 1 Relationship between CNR, C_{LB} and SD for each lesion size at the observer performance test.

CNR	C_{LB} [HU]	SD [HU]	C_{LB} [HU]	SD [HU]	C_{LB} [HU]	SD [HU]	C_{LB} [HU]	SD [HU]
Lesion diameter of 3.5 [mm]								
3.00	5	NA	10	NA	15	5.00	20	6.67
3.17		NA		NA		4.73		6.31
3.33		NA		NA		4.50		6.01
3.50		NA		NA		4.29		5.71
3.67		NA		NA		4.09		5.45
3.83		NA		NA		3.92		5.22
4.00		NA		NA		3.75		5.00
4.50		NA		NA		3.33		4.44
Lesion diameter of 5 [mm]								
1.50	5	3.33	10	6.67	15	10.00	20	13.33
1.67		2.99		5.99		8.98		11.98
1.83		2.73		5.46		8.20		10.93
2.00		2.50		5.00		7.50		10.00
2.17		2.30		4.61		6.91		9.22
2.33		2.15		4.29		6.44		8.58
2.50		2.00		4.00		6.00		8.00
3.00		1.67		3.33		5.00		6.67
Lesion diameter of 10 [mm]								
0.50	5	10.00	10	20.00	15	30.00	20	40.00
0.63		8.00		16.00		24.00		32.00
0.75		6.67		13.33		20.00		26.67
0.88		5.71		11.43		17.14		22.86
1.00		5.00		10.00		15.00		20.00
1.13		4.44		8.89		13.33		17.78
1.25		4.00		8.00		12.00		16.00
1.50		3.33		6.67		10.00		13.33
Lesion diameter of 15 [mm]								
0.40	5	12.50	10	25.00	15	37.50	20	50.00
0.50		10.00		20.00		30.00		40.00
0.60		8.33		16.67		25.00		33.33
0.70		7.14		14.29		21.43		28.57
0.80		6.25		12.50		18.75		25.00
0.90		5.56		11.11		16.67		22.22
1.00		5.00		10.00		15.00		20.00
1.10		4.55		9.09		13.64		18.18

CNR – contrast-to-noise ratio.

 C_{LB} – lesion-to-background contrast.

SD – image noise.

HU – hounsfield unit.

NA – not applicable.

Phantom study

Scanning the phantom

We obtained step-and-shoot- and helical scans on a 16 detector-row CT scanner (ECLOS, Hitachi Medical Corp., Chiba, Japan) using a low-contrast module of Catphan 600 (Phantom Laboratory, Cambridge, NY, USA) [16,17]. An external diameter of 200 mm, a contrast-difference level of 1.0% (CT number difference = 10 HU) and objects measuring 3.0–9.0 mm in diameter were used for visual evaluation. The CNR-based AEC technique was applied to phantom scanning. With CNR-based AEC, the tube current and rotation time were automatically assigned considering the size of the target lesion and the $C_{LB,Z}$. Before scanning, we set the target lesion size at 3.0–9.0 mm and the C_{PRE} value at 9 HU. Since C_{PRE} indicates an assumed C_{LB} under standard

conditions such as routine tube voltage and typical patient size, $C_{LB,Z}$ must become to be 10 HU at the scan position. In reference to Fig. 8, C_{COEF} equals approximately 1.11 at our experimental condition of 120 kV and 200 mm diameter. Therefore we set the C_{PRE} value at 9 HU. Note that if phantom external diameters are the same along z-axis position, $C_{LB,Z}$ are also the same. For the CNR-based AEC technique, 20 scans were performed. A total of 280 CT images were acquired by step-and-shoot and helical scanning.

The common scan parameters for step-and-shoot and helical scanning were: beam width at the center of rotation 10.0 mm, image thickness 5.0 mm, FOV 350 mm, tube voltage 120 kV, matrix 512×512 . The beam helical pitch was 0.94 at helical scanning, it is defined as the ratio between the table feed per rotation and the x-ray beam width. Images were reconstructed with the abdominal standard kernel (F32).

Visual evaluation of lesion detectability

We performed visual evaluation of lesion detectability. We extracted 65×65 pixel square areas that included low-contrast objects from the Catphan phantom images scanned with each CNR-based AEC technique. We placed 20 extracted images from 20 repeated scans for each lesion size in one image in Fig. 3 and used a total of 14 images in our visual evaluation of lesion detectability. The 14 images were generated using 14 different scanning conditions, i.e. 7 lesion sizes (3.0–9.0 mm) and 2 scan techniques (step-and-shoot scans and helical scans). The window width of 250 HU was set based on routine abdominal examination protocols and the window level, set at 50 HU, was adjusted to the Catphan background CT value. The 5 radiologists independently performed visual evaluations on an 8-bit grayscale display and recorded the number of lesions detected on 20 images. The number of confidently identified lesions was recorded; when the presence of a lesion was judged to be doubtful or denied with confidence, the extracted image was regarded as presenting no lesion. After visual evaluation of the images presented to the 5 readers we determined the detection fraction for each lesion size using the equation:

$$\text{detection fraction}[\%] = \frac{N_{\text{DetL}}}{20} * 100 \quad (3)$$

where N_{DetL} is the number of detected lesions. Finally, the mean detection fraction was calculated from the results reported by the 5 radiologists.

Results

Detectable CNR and lesion diameter

Averaged ROC curves for the 5 radiologists for all CNR values and lesion diameters were obtained (data not shown). The FPF value at TPF = 0.95 for each ROC curve is presented in Fig. 4 as a function of CNR for 4 lesion diameters. The solid lines in Fig. 4 show the fitted logistic

regression curves (Origin, version 7.5J; OriginLab, Northampton, MA) under 10 experimental conditions. The FPF values for each C_{LB} decreased as the CNR increased under all 4 lesion-diameter conditions. When the selection of the FPF value equals 0.05 for each regression curve, the dCNR value can be calculated.

Fig. 5 shows the dCNR values as a function of the lesion diameter at 3 different C_{LB} levels by integrating 2 C_{LB} conditions (5–10-, 10–15-, and 15–20 HU). The measured dCNR values are summarized in Table 2. For each C_{LB} condition, the dCNR derived from the observer performance test can be expressed by the power equation:

$$d\text{CNR} = a * D^b, \quad (4)$$

where a is a constant for each C_{LB} , D is the lesion diameter desired for detection, and b is the power factor. Here C_{LB} varies along with the water equivalent thickness in the same tube potential, so that equation (4) practically expresses as a function of z-axis position.

Fig. 5 indicates that a decrease in the lesion diameter or the C_{LB} requires a higher dCNR. Using equations (2), (4) and Fig. 5, SD_z is calculated at each z-axis position as shown in equation (5).

$$SD_z = k * D^l * C_{\text{LB},z}^m, \quad (5)$$

where k is a constant, l and m are power factors. In this paper, we set these values of k , l and m as $1.61\text{E-}2$, 1.26 and 1.52 , respectively. Since the equation (5) indicates a monotone increasing function, a decrease in the lesion diameter or the $C_{\text{LB},z}$ requires a lower image noise.

Visual evaluation of low-contrast detectability

With respect to low-contrast detectability, the mean detection fraction at the different lesion sizes is shown in Fig. 6 for step-and-shoot- and in Fig. 7 for helical scans. On step-and-shoot scans, the mean detection fraction with CNR-based AEC remained almost constant from 88 to 99% regardless of the lesion size in Fig. 6. We observed the same

Figure 3 Example of the visual evaluation of lesion detectability. Extracted images ($n = 20$) for each lesion size were placed in one image and 14 images (7 lesion sizes $\times$ 2 scan techniques) were used for the visual evaluation of lesion detectability.

Figure 4 FPF values as a function of CNR at TPF = 0.95 for 3.5- (a), 5- (b), 10- (c), and 15-mm (d) diameter lesions. Regression curves are shown as solid lines.

trend on helical scans, the mean detection fraction with CNR-based AEC exhibited a high score from 91 to 100% in Fig. 7.

The calculated $CTDI_{vol}$ and SD_z at each lesion size for the CNR-based AEC technique with $C_{LB,z}$ of 10 HU are summarized in Table 3. $CTDI_{vol}$ for the CNR-based AEC technique changed dramatically with the SD_z from 7.5 to 100.0 mGy for step-and-shoot scans and from 9.1 to 121.5 mGy for helical scans.

Figure 5 Graph showing the dCNR values as a function of the lesion diameter at 3 C_{LB} levels.

Discussion

CT is one of the most robust diagnostic modalities for detecting lesions [18]. Once the scan parameters such as kV and mAs are determined, lesion detectability depends on physical parameters such as C_{LB} and image noise. The SD -based AEC technique controls the tube current according to the image noise level predetermined on the CT image. However, from the standpoint of lesion detection, C_{LB} may more directly affect detection performance [19] than the image noise level. Therefore, we suggest that CNR is a more appropriate descriptor for controlling the radiation dose of AEC system. In this study, we elucidated the relationship between observer performance and the required CNR level for low-contrast lesion detection (i.e. dCNR). As a result, a decrease in the lesion diameter or the C_{LB} requires

Table 2 dCNR Values under 3 C_{LB} conditions obtained by integrating 2 C_{LB} conditions (5-10-, 10-15-, and 15-20 HU) for 4 lesion diameters (see also Fig. 6).

C_{LB} [HU]	Lesion diameter [mm]			
	3.5	5	10	15
5, 10	NA	3.03	1.02	0.56
10, 15	NA	2.48	0.78	0.46
15, 20	3.16	1.89	0.74	0.45

NA – not applicable.

Figure 6 Mean detection fraction at different lesion sizes: step-and-shoot scan. The error bars are standard deviation of mean.

a higher dCNR. Verdun et al. [17] reported the CNR thresholds allowing for 100% detection of low-contrast (8 HU) objects; the CNR required for the perfect detection of objects measuring 5-, 7-, and 9 mm in diameter was 1.0, 0.8 and 0.6, respectively. On the other hand, we found that when C_{LB} was 5 and 10 HU, the dCNR of 5-mm diameter lesions was 3.03, approximately three times that reported by Verdun et al. The difference between their and our results may be attributable to differences in the type of images used in the observer performance test, to manipulations of the image parameters, and to the window settings. Since our observers evaluated 8-bit grayscale images with 256 gray levels, the gray level corresponding to each CT number varies depending on image display parameters such as the window level and width (WL, WW). In general, the narrower the WW, the higher the detectability, and vice versa [20]. These results suggest that the displayed CNR is higher at a narrow- than a wide window width. Verdun et al. [17] set the WW at 100 HU, therefore

Figure 7 Mean detection fraction at different lesion sizes: helical scan. The error bars are standard deviation of mean.

Figure 8 Graph showing C_{COEF} values as a function of the water diameter at 3 tube voltages.

the dCNR of their 5-mm diameter lesions can be expected to be lower than in our study.

In addition, we investigated the low-contrast detectability using low-contrast (10 HU) module of the Catphan phantom. Low-contrast detectability is one of the most important factors in abdominal, especially hepatic CT [21]. On CT images, hepatic tumors are recognized by the tumor-liver contrast, in other words [22], C_{LB} . Previous study [8–10] reported that tumor-liver contrast was 5–40 HU in enhanced hepatic CT; however, in some cases, it was only a few Hounsfield units. Visual evaluation results revealed that CNR-based AEC improves the detection fraction of low contrast and small lesion at higher radiation dose.

Equation (2) indicates that a higher C_{COEF} or C_{LB} allows for higher image noise for lesion detection. In other words, low doses are allowed to maintain lesion detectability for high contrast or large size object. In a study of Yanaga et al. [23], the size of human hepatocellular carcinoma ranged from 7 to 44 mm and the tumor-liver contrast ranged from 21.3 to 53.2 HU although the injection duration and total iodine dose varied. High-contrast object scanning facilitates a reduction in the radiation dose while maintaining lesion detectability in CNR-based AEC. The smaller the C_{PRE} or the lesion diameter is input by user, the larger the radiation dose to the patient. We must set the well-considered predefined parameters of CNR-based AEC for every diagnostic task to balance between achieving the necessary level of image noise and keeping the radiation dose to the minimum.

Though the C_{COEF} values were higher when the tube voltage is lower as shown in Fig. 8, we did not evaluate the effects of tube voltage on lesion detectability or the radiation dose. Low tube voltage scanning may yield an advantage in CNR-based AEC in respect to the lesion detectability of radiation dose reduction. However, when a highly attenuated object such as a metal prosthesis is embedded in the body, severe beam-hardening and streaking artifacts are more prominent. Therefore, CNR-based AEC on lower tube potentials may not be preferred in such a region. We are planning additional studies on high contrast objects and low kV scanning techniques to gain further insights into the value of CNR-based AEC.

Table 3 Relationship between the calculated volume CT dose index ($CTDI_{vol}$) and SD_z at each lesion size for the CNR-based AEC technique with $C_{LB,z}$ of 10 HU.

Lesion diameter [mm]	Step-and-shoot scan			Helical scan		
	dCNRz	SD_z [HU]	$CTDI_{vol}$ [mGy]	dCNRz	SD_z [HU]	$CTDI_{vol}$ [mGy]
3	4.69	2.13	100.0	4.69	2.13	121.5
4	3.27	3.06	58.2	3.27	3.06	70.9
5	2.47	4.05	33.2	2.47	4.05	40.4
6	1.96	5.10	21.0	1.96	5.10	25.5
7	1.62	6.19	14.2	1.62	6.19	17.3
8	1.37	7.32	10.2	1.37	7.32	12.4
9	1.18	8.50	7.5	1.18	8.50	9.1

Note: To be accurate, water equivalent thickness at each z-axis position is not the same as 200 mm. Also, $C_{LB,z}$ is not the same as 10 HU. But we confirmed that calculated thicknesses were in error by less than 2 mm. Therefore, water equivalent thickness at each z-axis position and $C_{LB,z}$ of Table 3 are assumed to be 200 mm and 10 HU, respectively.

Our study has some limitations. First, in the visual evaluation experiment we presented simulated lesions that were hypo-dense compared to surrounding background structures. As Judy et al. [24] reported that hypo- tended to be more visible than hyper-contrast lesions, we are planning to carry out observer performance tests with hyper-contrast lesions. As shown in Fig. 5, the dCNR value of 3.5-mm lesions may exceed 4 when C_{LB} is less than 15. Under these conditions, the required image noise is below 3.75. If this parameter setting is applied to abdominal CT, higher tube currents are required to ensure adequate image quality. However, such higher tube currents cannot be achieved with currently-available CT systems due to limitations in the capacity of the x-ray tube. Second, the effect of other scanning parameters, such as FOV, reconstruction kernel and image thickness, on lesion detectability has not confirmed. The larger the FOV is chosen, the larger the pixel size. If the “abdominal standard” reconstruction kernel is used, images with reduced noise will be produced. Thicker slices improve contrast resolution and CNR. But spatial resolution will decrease in these cases. Lesion detectability will be affected not only by CNR but also by spatial resolution, therefore, the relation between dCNR and these scan parameters should be examined in the future. Finally, the dCNR values at 20 and greater C_{LB} remain uncertain. If CNR-based AEC is applied at hepatic multiphase CT, C_{LB} can be no lower than 20 and no higher than 100 during the abdominal phase [25]. Ideally, visual evaluation under expanded C_{LB} conditions should be possible.

In a final remark on the CNR-based AEC algorithm, we want to address both user input parameters and visual evaluation phantom. Before scanning, the AEC system demands that users input detection parameters of C_{PRE} and the desired lesion diameter. However, it may be difficult to determine these parameters ahead of CT study. Therefore, we intend to offer a solution that involves a new input parameter that allows easy determination of an appropriate image quality according to the diagnostic task.

In conclusion, we developed a CNR-based AEC algorithm and demonstrated the usefulness of the AEC technique. As CNR-based AEC adjusts the radiation dose for maintaining lesion detectability, it will yield great advantage in clinical practice.

Appendix

Determination of the conversion coefficient (C_{COEF})

We used a 16-detector-row CT scanner (ECLOS; Hitachi Medical Corp., Chiba, Japan). To measure the conversion coefficient, we constructed water-filled cylindrical phantoms with an internal diameter of 160-, 200-, 240-, 280-, and 320 mm and a 5 mm thick polymethylmethacrylate wall. These phantoms were scanned at tube voltages of 100-, 120-, and 130 kV at 400 mAs. A rod filled with an iodine-based contrast agent (Omnipaque; Daiichi Sankyo, Tokyo, Japan; 5 mgI/ml) or a rod filled with purified water was inserted at the central hole of the phantom. We reconstructed 2.5-mm-thick CT images using the abdominal standard kernel (F32). The CT number for each rod, the phantom diameter, and the tube voltage were averaged across 20 CT images. The water-to-iodine contrast was calculated as the difference between the CT number of the iodinated solution and the CT number of water. Fig. 8 shows the C_{COEF} as a function of the water diameter for 3 tube voltages. This value was normalized by the value of the water-to-iodine contrast at 120 kV and a water diameter of 280 mm. As in an earlier report [14], C_{COEF} was higher at lower tube voltages and decreased with increasing water diameter due to beam hardening. Therefore, lesion-to-background contrast at each z-axis position ($C_{LB,z}$) is calculated by utilizing the C_{COEF} as shown in equation (1).

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