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Research Article

Influence of the Display Field of View in Low-dose Computed Tomography on Airway Measurements

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Abstract

Purpose:
To examine the effects of reducing the display field of view (DFOV) in low-dose computed tomography on airway analysis of the chest.

Methods:
An airway phantom was created by embedding polytetrafluoroethylene tubes into the materials simulated the lung parenchyma. The phantom was scanned using the tube current-time products (mAs) setting from 5 to 150 mAs in two DFOVs (100 and 150 mm). The airway measurements were calculated as wall area ratios [%WA = wall area / (wall area + lumen area) × 100]. We used the volume CT dose indexes (CTDIvol) as indicators of the radiation dose. The airway measurements errors of different radiation doses were compared between the two DFOVs.

Results: The error of the airway measurements at low radiation doses in the 100-mm DFOV was lower than that at high radiation doses in the 150-mm DFOV based on the CTDIvol.

Conclusions: Airway measurements at low doses are most effective when the DFOV is reduced to 100 mm.

Keywords: Radiation Dose Reduction; Airway Measurements; CT; DFOV; Wall Area Ratios

Abbreviations:
CT: Computed Tomography;
CTDIvol: Volume Computed Tomography Dose Index;
DFOV: Display field of view;
HU: Hounsfield Units;
mAs: milliampere seconds (tube current-time products);
PTFE: Polytetrafluoroethylene;
SD: Standard Deviation ;
WA: Wall Area ;
%WA: Wall Area Ratio

Introduction

Chronic obstructive pulmonary disease, which is characterized by the presence of airflow limitation that is not fully reversible, includes small airway disease and emphysema [1]. The use of high-resolution volume data in the initial and follow-up assessment of patients with chronic obstructive pulmonary disease has facilitated airway measurements of the lung [2–4]. However, according to ALARA (As Low As Reasonably Achievable) principles, the radiation dose for the computed tomography (CT) protocol for airway analysis should be as low as possible for reasonably achieving effective imaging because radiation-induced carcinogenesis is a stochastic effect.

Dose reduction techniques have been reported [5], and low-dose CT has been shown to be comparable to standard-dose CT for airway bronchial measurements [6]. Airway measurements should be performed as accurately as possible. The measurement error was shown to be reduced when scanning was performed under a DFOV (or reconstruction field of view) of 200 mm [7, 8]. Furthermore, Rodriguez et al. found that using a half FOV (180 mm) with a higher frequency algorithm (bone) improved the accuracy of bronchial measurements at a low tube current-time products (mAs) setting (i.e., 12.5 mAs) [9]. However, the effectiveness and accuracy of measurements obtained at <12.5 mAs and a DFOV <180 mm have not yet been established.

Therefore, the purpose of this study was to examine the effects of reducing the DFOV during low-dose CT of the chest for airway analysis using a phantom model.

Materials and Methods

Phantom

Studies regarding the optimization of radiation dose in chest CT cannot be performed in human volunteers because of ethical limitations in conducting multiple CT scans of volunteers. Therefore, we used a phantom model in this study, which did not require ethical approval.

The object of the X-ray water phantom for chest and abdomen (JIS4915; Japan Medical Service Co., Ltd. Osaka, Japan [http://www.nihon-medical.jp/waterphantom/]) used in this study was configured with water, and the linear absorption and X-ray scattering and dispersion of the object was similar to the human body. When only the outer cylinder of the phantom was filled with water, the phantom was equivalent to a human chest, which is an X-ray absorber. For the airway phantom in our model (Figure 1a), the outer space of the X-ray water phantom for chest and abdomen was filled with water and the inner space was filled with mixtures of potato flakes and bread crumbs to simulate the lung parenchyma [10]; the polytetrafluoroethylene (PTFE) tubes were embedded in the inner space and simulated the airway wall. Furthermore, a water phantom (diameter, 67 mm) was embedded within the inner space. The PTFE tubes (external diameter, 5.04 mm; lumen diameter, 3.00 mm; length, 30 mm; wall thickness, 1.02 mm) were constructed to have 3 tilt angles (i.e., 0°, 30°, and 60°) from the z-axis of the CT couch in order to simulate the airways (Figure 1b).

The mean attenuation of the mixtures of potato flakes and breadcrumbs and the PTFE tubes was adjusted to approximately -850 Hounsfield units (HU) and 192 HU, respectively, as reported previously [8, 11].

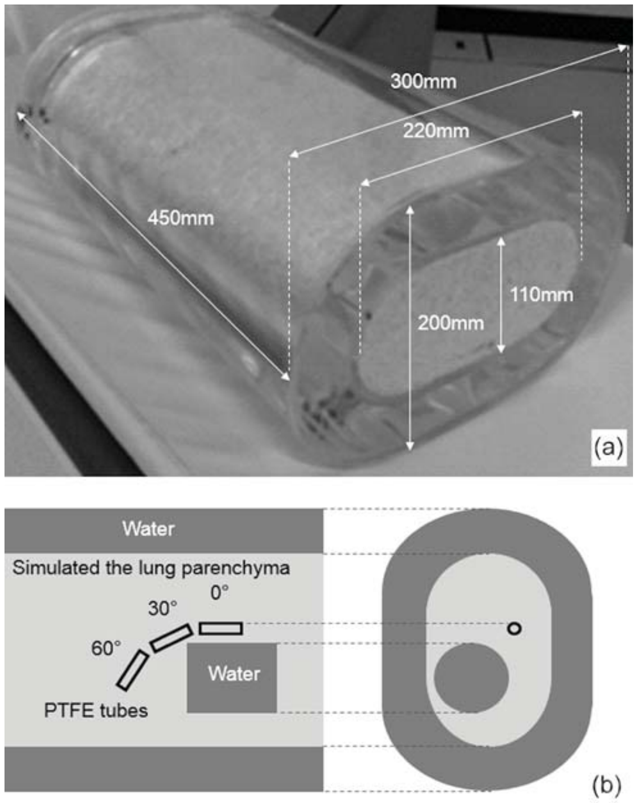

Figure 1. An airway phantom placed on the computed tomography (CT) scanner table. The (a) phantom sizes and (b) components of the phantom are shown.

CT imaging

The CT images of the airway phantom were obtained with a multidetector-row CT scanner (Aquilion 32; Toshiba Medical Systems, Otawara-shi, Japan) following a standard protocol for chest CT: 120 kV; rotation speed, 0.5 s; helical pitch, 0.719:1; scan FOV, 320 mm; DFOV, 100 and 150 mm; and beam width, 16 mm (0.5 mm × 32 detectors). All CT images were reconstructed in 0.5-mm–thick transverse sections at 0.3-mm intervals with a lung reconstruction kernel (FC50). CT images were obtained at 5, 10, 30, 50, 70, 90, and 150 mAs and were displayed with the following lung window settings: window width, 1500 HU; and window level, -600 HU.

Radiation dose

We used the volume CT dose index (CTDIvol; measured in mGy) as an indicator of the radiation dose during CT scanning. This parameter is a widely-accepted indicator of the radiation dose in CT scans [12].

Noise measurements

Noise data (i.e., the standard deviations [SDs] of CT values in a water phantom) were obtained at 5, 10, 30, 50, 70, 90, and 150 mA. The measurements were performed using a DFOV of 100 and 150 mm.

Noise was defined as three square regions of interest (1 central and 2 peripheral) in an area of 95–100 mm2 using Image J 1.48 (National Institutes of Health, MD, USA). The regions of interest that corresponded to the water phantom within the inner space were averaged over 10 slices (Figure 2).

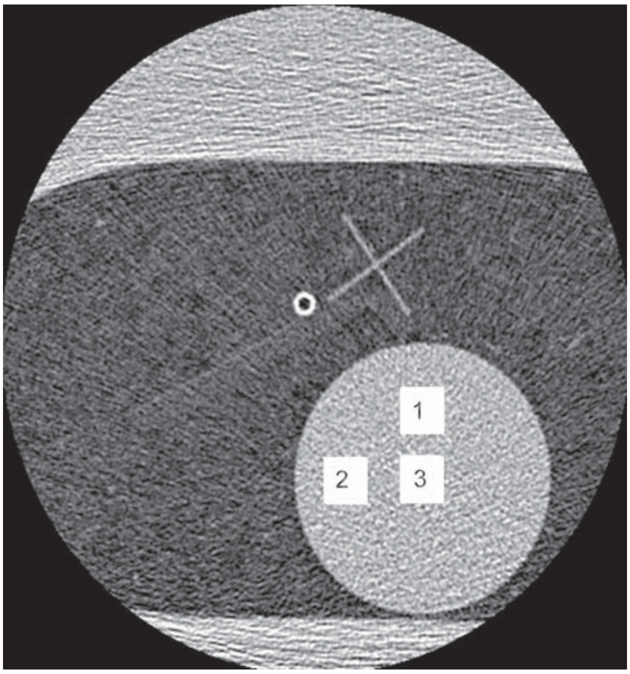

Figure 2. Cross-sectional images of 3 ROI placements. Noise measurement was considered the standard deviation of computed tomography values in a water phantom, and was measured in 3 square ROIs (1 central and 2 peripheral).

Airway measurements using automatic processing software

The reconstructed images were transferred to an image processing workstation (Advantage Work-station version 4.5, GE Healthcare, Milwaukee, WI, USA), and airway measurements were obtained using commercially available software (Thoracic VCAR, GE Healthcare, Milwaukee, WI, USA) in the airway analysis mode. We obtained values for the airway wall area (WA), lumen area, and wall area ratio (%WA). The %WA was defined as WA/(WA + lumen area) × 100. Thereafter, we assessed the mean %WA at 30 points along the tube. Measurement accuracy was assessed by the percent relative error of the phantom tube, which was determined as follows: percent relative error (%) = 100 × (measured value – actual value)/ actual value.

Airway measurements error and SD

The relationship between the airway measurement error (DFOV of 100 mm and a tilt angle of 0°) and SD was evaluated using the Spearman's rank correlation test; a p-value <0.05 was considered to indicate a significant difference. All statistical analyses were performed with the EZR software (version 1.11; Saitama Medical Center, Jichi Medical University, Saitama, Japan), a graphical user interface for R (version 2.13.0; The R Foundation for Statistical Computing, Vienna, Austria).

Results

Radiation dose

The CTDIvol for 5, 10, 30, 50, 70, 90, and 150 mAs were 0.8, 1.6, 3.8, 6.4, 8.9, 11.5, and 19.1 mGy, respectively.

Noise measurements

The results of the noise measurements from 5 to 150 mAs in two DFOVs (100 and 150 mm) are shown in Figure 3. The noise increased exponentially with increasing mAs.

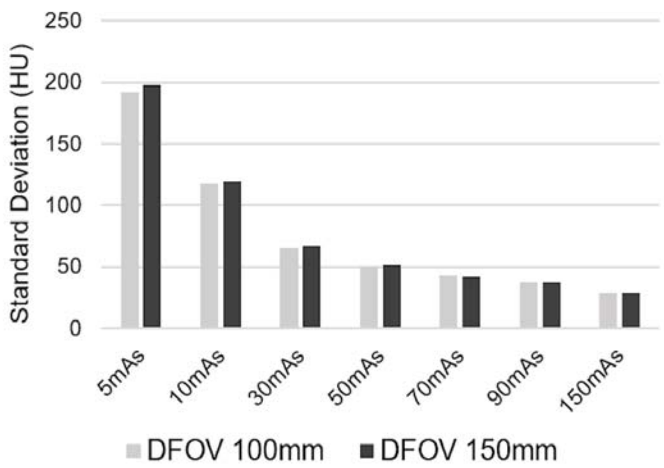

Figure 3. Noise measurements, defined as the standard deviation of computed tomography (CT) values in a water phantom, of CT images obtained at 5, 10, 30, 50, 70, 90, and 150 mA. The measurements were performed using a display field of view (DFOV) of 100 and 150 mm.

Measurement accuracy

The results of the percent relative error of the %WA at DFOVs of 100 and 150 mm for different mAs and tube angles for the airway phantom tubes are shown in Figures 4a and b, respectively. For all tube angles and mAs, the percent relative error of the %WA at DFOVs of 100 and 150 mm were <4% and <7%, respectively. The highest errors for the 100 and 150 mm DFOVs were observed at 5 and 10 mAs, respectively. Additionally, the tube angle influenced the error of the %WA for both DFOVs. The highest errors were observed at 5 mAs.

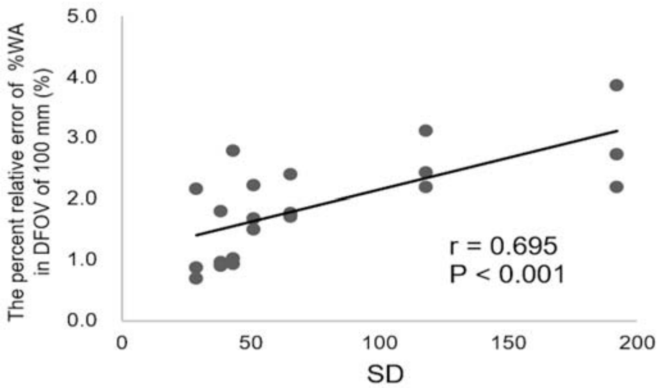

Figure 4. The correlation between the SD and the percent relative error of the %WA using a DFOV of 100 mm.

The percent relative error was increased at a tube angle of 60°. The results of the percent relative error of the %WA at DFOVs of 100 and 150 mm for different CTDIvol values and tube angles (0°, 30°, 60°) are shown in Figures 5a–c. The error at any dose and at a DFOV of 100 mm was smaller than the error at any dose at a DFOV of 150 mm.

Airway measurement error and SD

A strong correlation between the measurement error and SD at a DFOV of 100 mm was observed ($r = 0.695$, $p < 0.001$, Figure 6).

Discussion

According to the current study, the accuracy of airway measurements using the low radiation dose technique was improved with various algorithms and image reconstructions [6, 9, 13]. In a previous study, it was suggested that quantitative bronchial assessments preformed under low-dose CT (25 mAs) can be potentially substituted for standard-dose CT (150 mAs) [6]. Similar to this previous study, we were also able to show that variations in the errors decreased from 30 to 150 mAs at both DFOVs. However, to our knowledge, studies involving bronchial measurements <12.5 mAs at small DFOVs have not yet been conducted.

In this study, the changes in the variations of errors at low-dose from 5 to 10 mAs were larger compared to that at 150 mAs. This difference was considered to be influenced by noise. Therefore, we investigated the relationship between noise and measurement errors in multiple doses. More prominent effects of the noise measurements (Figure 3) were observed at a SD of 5 and 10 mAs. Joemai et al. also showed that noise measurements at 10 mAs were different compared to those at other mAs settings because there was a proportionally large effect of noise in relation to the low dose [14].

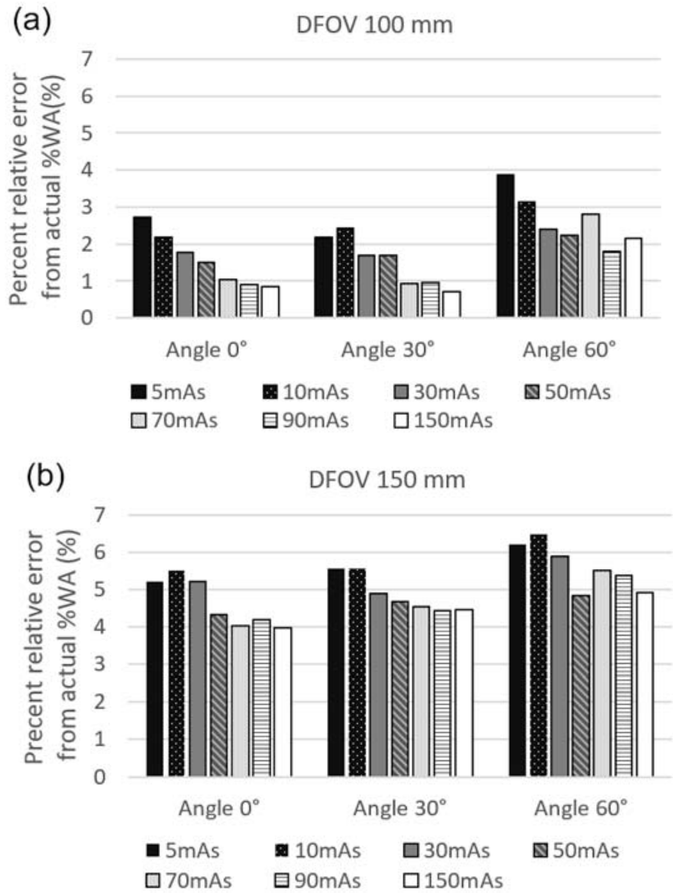

Figure 5. The percent relative error of the WA for different tube current-time products settings (mAs) at 3 tube angles using a DFOV of (a) 100 mm and (b) 150 mm.

Therefore, in this study, the relationship between SD and the error was highly correlated, which could explain why the errors in airway measurements were larger due to the influence of rapidly increasing SD values during the low-dose setting. Therefore, we considered that the effects in the variation of errors from 30 to 150 mAs were small because there was a smaller increase in the SD values. In this software (Thoracic VCAR, GE Healthcare, Milwaukee, WI, USA), the detection of airway and measurements were conducted using watershed segmentation and region growing algorithms. In a previous study, it was demonstrated that a noise with a high SD led to a large error ratio when detecting the segmentation and boundary in a binary image using the region growing approach for watershed segmentation [15].

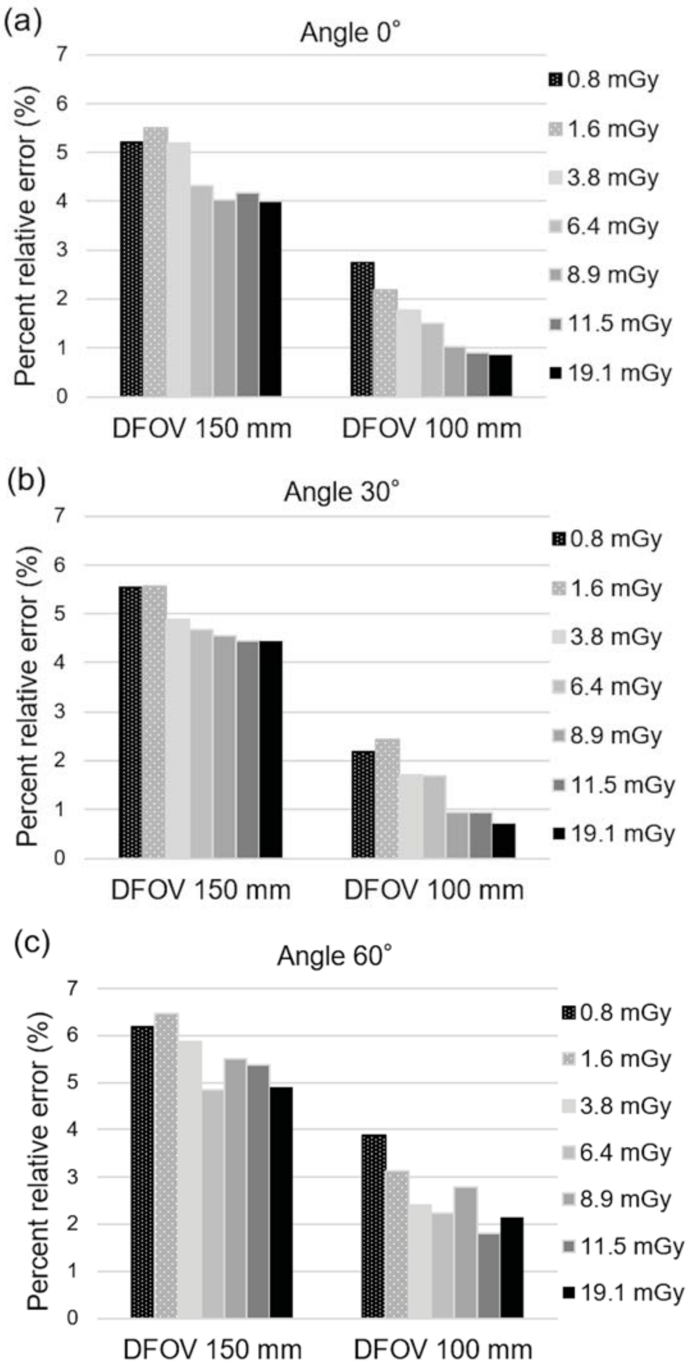

Figure 6. The percent relative error of the wall area ratio using a display field of view (DFOV) of 100 and 150 mm for different volume computed tomography dose indexes obtained at tube angles of (a) 0° (b) 30°, and (c) 60°.

Previous researchers have reported that using a small DFOV improved quantitative airway measurements [7, 9, 16]. The accuracy of measurements improved after using the DFOV (18 cm) while varying the reconstruction kernels and algorithms [9]. Our results indicated that a small DFOV improved airway measurements. The relative error for the 150-mm DFOV ranged 4.0–6.5%, and was significantly lowered (0.7–3.9%) when the DFOV was reduced to 100 mm. Takahashi et al. [7] showed that the error ratio was small for a DFOV of 200 mm with a 1.0-mm wall thickness, and they recommended that CT images should be obtained at a DFOV of approximately 200 mm.

According to a previous report [16], a large DFOV of 360 mm in matrix values of 512 × 512 (large pixel size 0.703 × 0.703) carries a considerably larger relative error than a small DFOV (100 mm, 200 mm) at wall thickness of 1 mm, and a DFOV of 100 mm in matrix values of 512 × 512 (pixel size 0.195 × 0.195) carried a larger relative error than 150-mm DFOV. That report also suggested that the voxel size should be reduced to 0.195 × 0.195 × 0.8 mm to analyze the bronchial airways with higher accuracy at DFOVs of 50–360 mm on a matrix of 512 × 512 or 1024 × 1024. In this study, we obtained similar results: the relative error of the %WA at reconstructed DFOVs of 100 and 150 mm (on a matrix of 512 × 512) was <4% and <7%, respectively, which allowed for error improvement in a small DFOV of 100 mm (pixel size 0.195 × 0.195). Achenbach et al. [16] investigated the combination of DFOV and matrix settings, which improved image quality and increased visualization of the bronchial wall at small pixel sizes, and resulted in a step-like morphology of the bronchial wall at large pixel sizes (0.703 × 0.703 mm, 360-mm DFOV, and 512 × 512 matrix). We considered that the voxel size used in this study was appropriate. Therefore, the relative error of %WA suggested that an improvement in the accuracy of airway measurements at a DFOV of 100 mm could be obtained by using optimized voxel sizes and small FOV.

In general, it is well known that there is a linear relationship between tube current and image noise due to an increase in the number of photons. It is also well known that a small DFOV increases image noise while improving spatial resolution [17]. However, it has been shown that the accuracy of measurements becomes worse with full DFOV (36 cm) [9]. Similar results have been obtained with CT airway measurements [7, 16]. Therefore, we tested only two DFOVs (100 mm and 150 mm), which were no larger than 150 mm. Moreover, we could not perform airway measurements at DFOV <100 mm using current software.

Sato et al. [18] investigated the effects of reconstruction FOV and reconstruction kernel on aliased noise. The authors found that large FOVs increased the noise and that a reconstruction FOV of 20 cm was an alias-free state. In addition, they estimated that the noise power spectrum in the FC13 kernel option was higher at a large FOV and that noise was higher in FC30 than in FC13.Yukimura et al. [17] recently reported that SD depends on DFOV. Together, these two studies have shown that noise changes with kernel at large DFOVs, but not at small DFOVs.

In this study, a voxel size of 0.195 × 0.195 × 0.5 mm reduced the error. When matrix values were 512 × 512 in the reconstructed 100-mm DFOV and 150-mm DFOV, the reconstructed pixel size was 0.195 mm × 0.195 mm and 0.293 mm × 0.293 mm, respectively. When matrix values were 512 × 512 in the scanned 320-mm FOV, the scanning pixel size was 0.625 mm × 0.625 mm.

Our study included an evaluation of SD by reconstructed FOV that might explain the reduced image noise. SD values were similar at any dose, both at 100-mm and 150-mm DFOV (Fig. 3), because small DFOVs (100 mm and 150 mm) produced the least noise under an alias status [18], and the kernel exerted no effect [17]. Furthermore, the errors at low dose were large due to the influence of high noise, and small DFOV improved airway measurements. For this reason, in airway measurements with a small DFOV, we consider that improved image quality in small pixel sizes is more influential than noise.

To the best of our knowledge, there were no studies conducted regarding the estimation of low dose at a DFOV <150 mm. In the airway measurements, the doses should be as low as possible because the CT scan is repeatedly performed for follow-up. Furthermore, measurement errors should also be as low as possible. Therefore, we assessed the influence of the DFOV in the resulting percent relative error for different CTDIvol values. Furthermore, different CTDIvol values were used to compare measurement errors in low absorbed radiation doses. Errors at low radiation doses (0.8 mGy) and a DFOV of 100 mm were smaller than those at standard radiation doses (1.6 mGy, 3.8 mGy) and a DFOV of 150 mm. According to our findings, reducing the DFOV to 100 mm is important for conducting airway measurements using a low dose.

We evaluated various tilt angles in the phantom placed from the z-axis of the CT couch, because the angle of airway geometry in a human tracheobronchial tree should be considered. Measurements at large tilt angles have been reported to result in large errors [11]. The influence of the tilt of the airway must be considered if the tube wall is <1.5-mm thick [7]. Indeed, our study results supported this conclusion. Therefore, it is necessary to use caution when evaluating the bronchus at various angles.

Limitations

The lung phantom and PTFE tubes used in this study to simulate the airways do not completely resemble the lung parenchyma and airways in humans. However, airway measurements using phantoms have been performed widely for previous investigations [7, 8, 9, 13]. Saba et al. [19] used an airway phantom embedded in potato flakes (approximately -650 HU) to simulate the lung parenchyma. A portion of the lung with less damage from emphysema changed the attenuation between -850 and -950 HU. The current threshold point of general pulmonary emphysema is -950 HU. However, -960 HU would be a better cut-off value according to recent data [20]. In recent years, the CTP675 lung phantom (The Phantom Laboratory, Inc., Salem, NY, USA) is widely used in bronchial measurements; the mean attenuation of the simulated lung parenchyma was -856 HU. Therefore, the mean density of our lung phantom was adjusted to approximately -850 HU.

Tubes of different materials with physical densities of 0.9–2.1 g/cm3 have been previously reported for stimulating the airway [8]. Weinheimer et al. [21] obtained a CT number of about 207 HU using tubes with a physical density of 1.14 g/cm3. Therefore, we also used tubes with a similar physical density. In this study, we assessed the wall thickness of only 1.0 mm based on Nyquist's theorem, which states that a wall thickness measurement of 1.0 mm requires a spatial resolution of <0.5 mm [22]. In several studies, wall thickness measurements of 0.5–1.0 mm are not considered accurate, whereas measurements >1.5 mm are considered to have more precision [7, 11]. In a previous report [7], the error ratio was small for DFOVs of ≤200 mm with a 1.0-mm wall thickness.

The use of different sizes of tubes has been previously reported [8, 9]. The error in the %WA was about 50% when using tubes with a wall thickness of 0.5 mm [8]. Furthermore, Rodriguez et al. [9] determined that the error in the %WA for a tube with a wall thickness of 0.4 mm was as high as about 80%. We did not use a wall thickness <1 mm in this study, which could result in large errors when analyzing the error in the %WA according to the dose and DFOV more accurately.

In our analysis, we only used the lung kernel because several researchers have found that the errors in airway measurements obtained using different CT scanners were large when a softer kernel was used and small when the bone and lung kernel were used [8, 9, 23].

It is not possible to use iterative reconstruction methods on the CT system used in this study. However, iterative methods are important in low-dose CT because they improve noisy images. Therefore, it would be worth to focus on the use of iterative methods of image reconstruction based on sparse sampling theory, such as total variation or other more sophisticated approaches, enabling an additional decrease of the radiation dose by reducing the angle or the number of projections [24]. Iterative methods might be indispensable for airway measurements in the future.

Conclusion

The accuracy of airway dimensions measured using CT images was affected by the DFOV and pixel size. Using a small DFOV improved the airway measurement errors. Reducing the DFOV to 100 mm is very important to obtain airway measurements at low radiation doses. In clinical bronchial measurements, it is expected that reducing the DFOV to 100 mm can reduce exposure.

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Conflict of Interest

The authors declare that they have no conflicts of interest.

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