An improvement of the computational effective diameter measurement in thoracic computed tomography examinations

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ABSTRACT

A method to calculate a corrected effective diameter (D_{MIL}) to more accurately estimate the dose received by a patient in chest computed tomography (CT) examination had been previously proposed. However, the discrepancy between D_{MIL} and water-equivalent diameter (D_w) is still relatively high (i.e. about 6%). Furthermore, the method is still performed manually, so it is laborious and time-consuming. This study aims to improve the corrected effective diameter with bone correction (D_{eff}^{corr}) and to automatically calculate it. The automated D_{eff}^{corr} was calculated as the square root of the product of these corrected AP and LAT diameters. The approach was implemented on 30 patients who had undergone chest CT examination with the standard protocol. The results show that the correlation between the D_{eff}^{corr} and D_w is $R^2=0.93$ with no statistical difference ($p>0.05$). The automated D_{eff}^{corr} is 3.1% lower than D_w. While the D_{MIL} is 10.5% higher than D_w and both are statistically different ($p<0.05$). In conclusion, the new D_{eff}^{corr} was introduced and the result obtained was closer to D_w than D_{MIL}. This method is simple enough to be used as an alternative method to accurately estimate D_w for radiation dose estimation in clinical chest CT scanning.

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1. INTRODUCTION

Computed tomography (CT) is one of the medical imaging modalities widely for early diagnosis of disease [1], [2]. Although it has enormous benefits, CT delivers relatively high radiation dose to the patient [3], [4]. It was reported that the dose received by patients in CT examination is about 15 mSv in an adult and 30 mSv in neonates for every single CT examination [5], whereas examinations with mammography, conventional X-ray, and nuclear medicine deliver radiation doses of 0.3-0.6 mSv, 5-10 mSv, and 2-5 mSv, respectively [6]. Moreover, the dose delivered by a single CT examination is greater than the annual radiation dose received from natural radiation sources such as radon and cosmic radiations (1-10 mSv) [7].

The United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR) reported that high radiation doses pose a higher risk of cancer in the general population [8]. Consequently, the topic of CT dosimetry is always of high interest [9], [10]. Radiation dose monitoring in CT examination is still based on the volumetric computed tomography dose index ($CTDI_{vol}$) and the dose-length product (DLP). However, these dose indices do not consider the size of patients [11]. It was reported that for the same $CTDI_{vol}$ value, the patient dose increases with decreasing patient size [12], [13]. Therefore, although the $CTDI_{vol}$ and DLP indices are accurate in estimating the radiation output of a CT scanner, they are less accurate in assessing the dose received by the patient [14], [15]. The American Association of Physicists in Medicine (AAPM) in 2011 introduced another dose index that considers the patient size, the size-specific dose estimates (SSDE) [16]. The size of a patient is initially represented by the effective diameter (D_{eff}) [15]. However, it is known that D_{eff} does not consider the different attenuation properties of the various body parts. Differences in attenuation affect the absorption of radiation which impacts the dose received by the patient [17].

Water-equivalent diameter (D_w) was subsequently introduced by AAPM in task group 220 to incorporate both the size and attenuation properties of a patient's body [18]. Automated methods to obtain the value of D_w have been proposed by many researchers recently, such as Anam *et al*. [19], Özoyka *et al*. [20], Gharbi *et al*. [21], and Juszczyk *et al*. [22]. Calculation of D_w needs specific software since it involves calculation of the average CT number and the area of the patient's image [23]. Not all hospitals have such software so an alternative method for estimating D_w is needed.

Mihailidis *et al*. [24] introduced a method to calculate the value of corrected effective diameter as an alternative to calculate the D_w in thoracic CT examinations. The lateral (LAT) thickness is corrected by the average relative electron density of the lung (ρ_e=0.3) to overcome the presence of the lung, so that the results obtained are close to the value of D_w. However, the values obtained from this method are significantly different (i.e. about 6%) from the value of D_w because the presence of bone is not considered. Furthermore, this method is carried out manually using an electronic ruler and involves many steps. The purpose of this study is to improve the effective diameter correction with both lung and bone to obtain the result closer to the D_w and to propose an automated method for calculating it.

2. METHOD

2.1. The diameters calculation

If we assume that the cross section of the patients is circular or elliptical, we can estimate the effective diameter of patients from the magnitude of the diameters in the anterior-posterior (AP) and lateral (LAT) directions (1).

$$D_{eff} = \sqrt{AP \times LAT} \tag{1}$$

But, in fact, the patient's cross section is neither elliptical or circular [25]. In addition, D_{eff} does not consider the attenuation properties of body.

The AAPM proposed the water-equivalent diameter (D_w) to incorporate both the size and attenuation properties of the patient's body [18], as seen in (2).

$$D_w = 2 \times \sqrt{\left(\frac{ROI_{mean}}{10000} + 1\right)} \times \sqrt{\frac{A_{ROI}}{\pi}} \tag{2}$$

Where ROI_{mean} is the mean CT number of the region of interest and A_{ROI} is total area of ROI of the axial image. In the D_w calculation, specific software is needed to calculate the average CT number and the area of the patient's image.

Mihailidis *et al.* [24] proposed a different approach for considering the lung tissue in the image. They corrected the lateral diameter with the relative electron density of lung tissue (ρ_e=0.3), naming it the LAT_{eff}^{corr}, as seen in (3). It was hoped that this value would be equivalent to D_w [24].

$$D_{MIL} = \sqrt{AP . LAT_{eff}^{corr}} \tag{3}$$

We postulate that to achieve more accurate results, the correction should not only be made in the LAT direction but also in the AP direction. Correction must account for the presence of bone as well as the presence of lung tissue. Therefore, we propose the corrected effective diameter (D_{eff}^{corr}) as seen in (4).

$$D_{eff}^{corr} = \sqrt{AP_{eff}^{corr} . LAT_{eff}^{corr}} \tag{4}$$

In addition, we propose an algorithm to automate the calculation of D_{eff}^{corr} to make it easier to implement in a clinical setting. The steps are shown in Figure 1, while the example from patient image is shown in Figure 2. First, we open the original image as seen in Figure 2(a). Then, we segmented the patient border automatically and converted the CT values outside the patient to a pixel value of 0 Figure 2(b). Then we detected the presence of lung and bone. The lung was detected with a threshold of -500 HU and bone with a threshold of +100 HU. CT values lower than -500 HU were converted to a pixel value of 1. CT values in the range -500 HU≤x≤+100 HU were converted to a pixel value of 2. All other CT values were converted to a pixel value of 3. We then determined the center position of the patient and determined the diameters in AP and LAT from the central image Figure 2(c). In the lung tissue (i.e., with pixel value of 1), the AP and LAT diameters were corrected with the average relative electron density of lung (ρ_e=0.3), and in the bone (i.e., pixel value of 3) corrected with the average relative electron density of bone (ρ_e=1.8). Finally, the D_{eff}^{corr} was calculated using (4) and as seen in Figure 2(d). The algorithm was effectively integrated into a graphical user interface (GUI) of IndoseCT 20b as depicted in Figure 3.

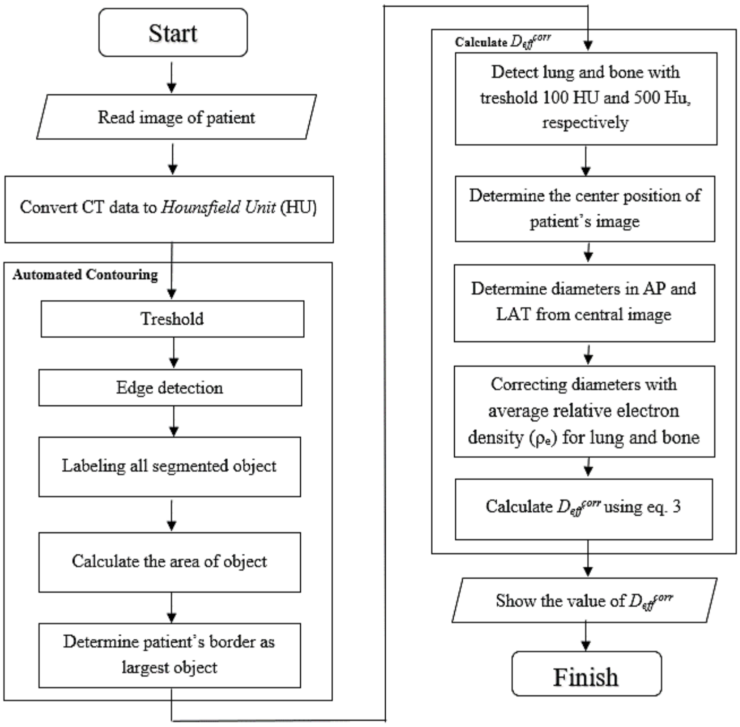


Figure 1. Flowchart for automatic corrected effective diameter (D_{eff}^{corr}) calculation

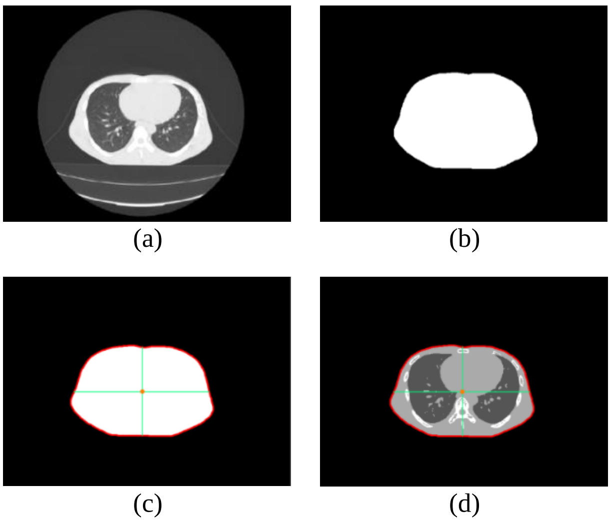


Figure 2. The segmentation process of D_{eff}^{corr}, (a) original image of patient, (b) the image after binarization, (c) the central position of LAT and AP diameters to estimate the central effective diameter (D_{eff}), and (d) Result of patient image with changes in lung pixel value to 1, bone to 3, and otherwise to 2

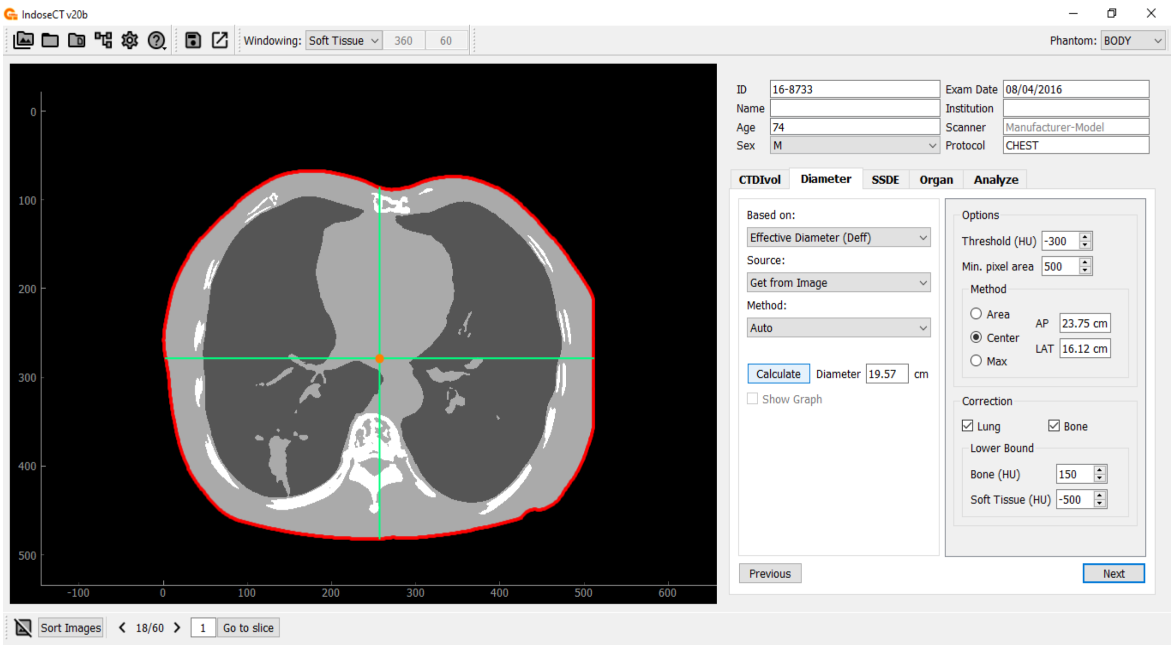


Figure 3. The screenshots of IndoseCT 20b for an automatic calculation of corrected effective diameter in thoracic region

2.2. Patient images

In this study, we evaluated 30 patients who undergone chest CT examination. The examinations were performed using a 128-slice multi-detector CT scanner, the Toshiba Aquilion 128. The patients were scanned with standard imaging protocol using a voltage of 120 kVp, 3D tube current modulation (TCM), a pitch of 1,438 and a collimation beam width of 64×0.5 mm in Ken Saras Hospital Semarang, Indonesia.

3. RESULTS AND DISCUSSION

The results of the calculated diameters (D_{MIL}, D_{eff}^{corr}, and D_w) are shown in Figure 4. The average and standard deviation of the diameters (D_{eff}^{corr}, D_{MIL}, and D_w) are listed in Table 1. The D_{eff}^{corr} is slightly smaller than D_w. The percentage differences between D_{MIL}, D_{eff}^{corr}, and D_w are shown in Table 2. The differences between D_{eff}^{corr} and D_w is 3.1 % and they are not statistically significant (p>0.05). However, D_{MIL} is 10.5 % different from D_w. The D_{MIL} is statistically different (p<0.05) from D_w.

The relationship between the diameters is shown in Figure 5, with the automated D_w and D_{eff}^{corr} in Figure 5(a), and the automated D_w and the D_{MIL} in Figure 5(b). The automated D_{eff}^{corr} is linearly correlated (R^2=0.93) with D_w. The correlation between D_w and D_{MIL} is R^2=0.95.

The purpose of this study was to improve the corrected effective diameter with bone correction (D_{eff}^{corr}) and to make it is automatically carried out. Previously, Mihailidis *et al*. [24] proposed a manually corrected D_{eff} (D_{MIL}) and the results were expected to close to D_w. However, their results were still 6.0 different from D_w [24] because the correction was only performed in the LAT direction and for lung only. Furthermore, their method is laborious and time-consuming. We have proposed corrections in two directions (AP and LAT) and inclusion of bone in addition to lung. We subsequently develop a software to automatically calculate the corrected D_{eff} (D_{eff}^{corr}) which works quickly and. The corrections are crucial to obtain a value of D_{eff}^{corr} close to D_w so that the calculation of SSDE is more accurate.

In Table 2, the value of automated D_{eff}^{corr} is only slightly lower (3.1%) than the value of D_w. This is because the calculation of D_{eff}^{corr} considers lung and bone of the thorax. The presence of lung and bone in thorax are corrected with the average relative electron density of lung (ρ_e lung=0.3) and bone (ρ_e bone=1.8) in both the LAT and AP directions. In this study, the D_{MIL} is 10.48% lower than D_w. The larger difference between D_{MIL} and D_w is because D_{MIL} corrected the image with the attenuation of the lung in only one direction (only in the LAT diameter), while the value of D_w was corrected in two directions [24]. Ignoring bone in AP dimension in the D_{MIL} method made the values obtained far lower than D_w.

The main finding of our study is that the automated corrected effective diameter can be close to water-equivalent diameter if we use a correction for the AP diameter as well as considering the presence of bone. The automated D_{eff}^{corr} can be used as a viable alternative since it has the close value to D_w. However, automatic detection of lung and bone can be tricky because it depends on the pixel thresholds used. In this study, we used -500 HU to detect the lung and +100 HU to detect the bone. The thresholds maybe different for other images.

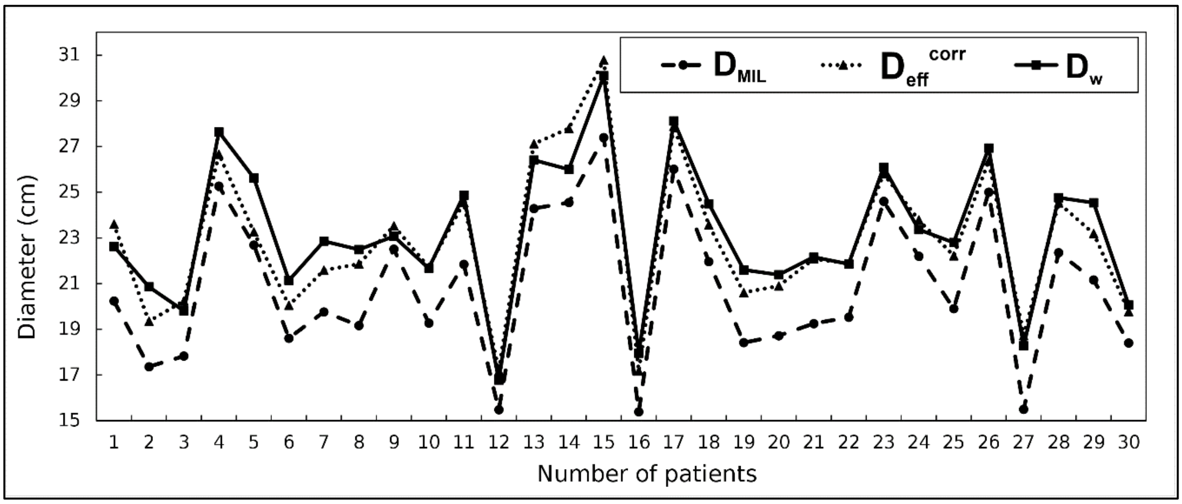


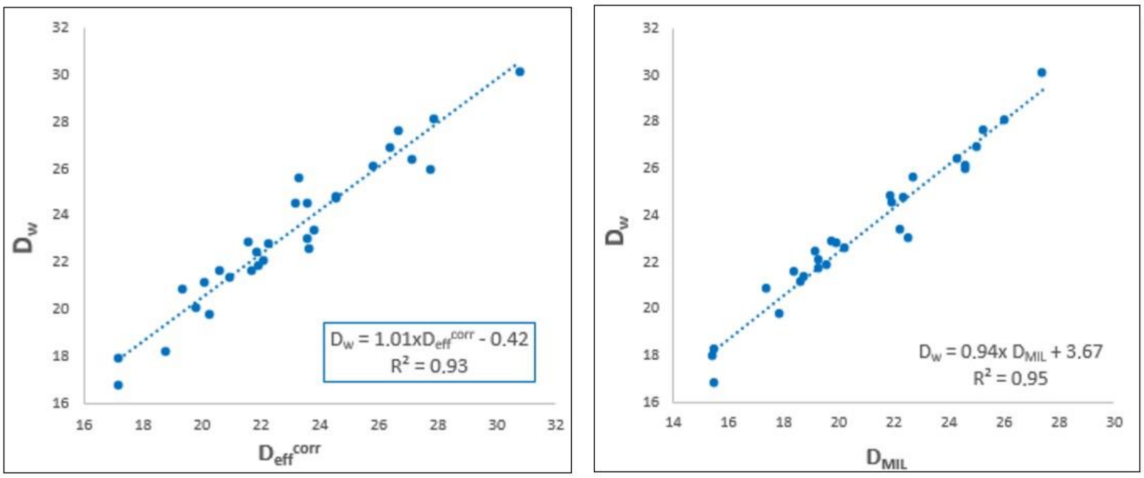
Figure 4. Results of D_{eff}^{corr}, D_{MIL}, and D_w in thoracic examinations

Table 1. The D_{MIL}, D_{eff}^{corr}, and D_w values

Parameter	D_{MIL}	D_{eff}^{corr}	D_w
Mean (cm)	20.82	22.93	23.21
Std deviation (cm)	3.22	3.37	3.21
Max (cm)	27.39	30.78	30.1
Min (cm)	15.39	17.17	16.78

Table 2. Percentage differences the D_{eff}^{corr} and D_{MIL} from D_w

Parameter	Mean	Std Deviation	Max	Min	*p* value
D_{eff}^{corr} (%)	3.06	3.61	9.21	0.09	0.72
D_{MIL} (%)	10.48	2.22	16.76	2.42	<0.01



(a) (b)

Figure 5. The relationships between (a) D_{MIL} and D_w, and (b) D_{eff}^{corr} and D_w in thoracic examinations

4. CONCLUSION

The new D_{eff}^{corr} was introduced and the result obtained was closer to D_w than D_{MIL}. A software to automatically estimate D_{eff}^{corr} has been developed and tested. It can be operated easily and quickly. The corrected effective diameter gives the close result to D_w (i.e. the difference is about 3%) and it can be an alternative to measuring D_w in daily examinations in routine clinical chest CT.

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