Use of Water Equivalent Diameter for Calculating Patient Size and Size-Specific Dose Estimates (SSDE) in CT:

The Report of AAPM Task Group 220

Cynthia McCollough, Ph.D., FAAPM[1] [Chairperson], **Donovan M. Bakalyar, Ph.D.[2]**, **Maryam Bostani, Ph.D.[3]**, **Samuel Brady, Ph.D.[4]**, **Kristen Boedeker, Ph.D.[5]**, **John M. Boone, Ph.D., FAAPM[6]**, **H. Heather Chen-Mayer, Ph.D.[7]**, **Olav I. Christianson, Ph.D.[8]**, **Shuai Leng, Ph.D.[1]**, **Baojun Li, Ph.D.[9]**, **Michael F. McNitt-Gray, Ph.D., FAAPM[3]**, **Roy A. Nilsen, B.S.[10]**, **Mark P. Supanich, Ph.D.[11]**, and **Jia Wang, Ph.D.[12]**

[1]Mayo Clinic, Rochester, MN

[2]Henry Ford Health System, Detroit, MI

[3]UCLA Medical Center, Los Angeles, CA

[4]St. Jude Children's Research Hospital, Memphis, TN

[5]Toshiba Medical Research Institute, Los Angeles, CA

[6]UC Davis Medical Center, Sacramento, CA

[7]National Institute of Standards and Technology, Gaithersburg, MD

[8]Landauer Medical Physics, Allison Park, PA

[9]Boston University School of Medicine, Boston, MA

[10]GE HealthCare, Waukesha, WI

[11]Rush University Medical Center, Chicago, IL

[12]Stanford University, Stanford, CA

1. Introduction

Volume computed tomography dose index ($CTDI_{vol}$) and dose length product (DLP) values are frequently used to represent radiation doses from a CT scan. The limitation of $CTDI_{vol}$ and DLP is that they are surrogates for patient dose, providing information about the scanner output for only a very specific standardized condition [1]. The dose received by a patient depends on both patient size and scanner output. Turner et al. showed that using $CTDI_{vol}$ as a normalization factor, organ dose estimates can be obtained for a specific patient size [2], and that the relationship to patient size was consistent across scanner models [3]. Their results showed a promising approach to estimating size-dependent, scanner-specific, and

exam-specific organ doses based on patient size and the $CTDI_{vol}$ reported by the scanner. Hence, obtaining accurate information about patient size is crucial to estimating patient dose in CT.

Report 204 from the American Association of Physicists in Medicine (AAPM) described the use of a size metric that involved the physical dimensions of the patient (anteroposterior [AP], lateral, AP+lateral, or effective diameter), in combination with scanner output ($CTDI_{vol}$), to determine size-specific dose estimates (SSDE) from CT scanning [4]. Patient dimension can be determined using physical or electronic tools. Physical devices, such as the calipers that were frequently used in radiography before the routine use of phototiming, may be used to measure patient thickness in the AP or lateral directions. Alternatively, electronic measurement tools can be used to measure physical dimensions from either the CT localizer radiograph or an axial CT image. The conversion factors used to calculate SSDE from $CTDI_{vol}$ reported in AAPM Report 204 were derived from experimental and Monte Carlo data and normalized to patient size in terms of water- or tissue-equivalent materials.

For the task of calculating SSDE, geometric size was used as a surrogate for a patient's x-ray attenuation. However, x-ray attenuation is the fundamental physical parameter affecting the absorption of x-rays and is thus more relevant than geometric patient size in determining the radiation dose absorbed by the patient. For example, regions of the thorax and abdomen could have the same external physical dimensions. However, because the lungs are less dense and of different composition than abdominal tissue, the thorax would attenuate fewer x-ray photons than would the abdomen. For the same scanner output ($CTDI_{vol}$), the thorax region would experience a higher radiation fluence and, hence, have a higher absorbed dose than an abdominal region having the same geometric dimensions. While CT operators can measure a patient's AP or lateral width, they currently have no practical way to measure attenuation. Both a CT localizer radiograph and CT projection data are measurements of the integrated x-ray attenuation along a ray path, and a CT image is a cross-sectional map of the linear attenuation coefficients of the materials in the image, normalized to the linear attenuation of water. Therefore, the CT localizer radiograph, the CT projection data, and the CT image all contain information that can be used to estimate patient attenuation.

The charge of AAPM Task Group 220 was to develop a robust and scientifically sound metric for automatically estimating patient size in CT that would account for patient attenuation and allow routine determination of SSDE for all patients, with little or no user intervention. This task group had a specific goal of developing a practical, standardized approach to estimating patient size that could be implemented by CT scanner manufacturers and others using CT localizer radiographs, axial CT images, or other data derived from the scanning process (e.g., projection data). Advantages and limitations of different methods were considered during task group deliberations and are summarized in this report. This includes comparing the SSDE calculated using various geometric size metrics, such as AP or lateral dimensions and effective diameter, and attenuation metrics, such as water equivalent diameter. Finally, recommendations are presented on the adoption and implementation of a standardized approach to estimating patient size.

This report is organized as follows. First, the concept of water equivalent diameter (D_w) is presented, and the methodology of calculating it from either a CT image or a CT localizer radiograph image is described. Second, data are provided comparing the accuracy of Monte Carlo dose estimates made using geometrical-based versus attenuation-based metrics for a series of virtual abdomen and thorax phantoms and their respective virtual CT images, and for patient images. Third, data are provided comparing D_w calculations from CT image and CT localizer radiograph phantom measurements. Fourth, practical considerations involved in implementing either approach are discussed, and recommendations for users and for manufacturers are provided. Finally, a road map for commercial adoption is suggested such that both patient size and SSDE can be calculated in a robust and consistent fashion across CT scanner manufacturers, and the resultant values stored in either the DICOM image header or the DICOM-structured dose report.

2. Water Equivalent Diameter (D_w)

Previous work has expressed the x-ray attenuation of a patient in terms of a water cylinder having the same x-ray absorption [5-9]. The area and diameter of such a cylinder of water are referred to as the water equivalent area (A_w) and water equivalent diameter (D_w), respectively. While the concepts of water equivalent area and diameter were proposed in these studies, doses to the scanned object and its water equivalent cylinder were not actually compared in order to demonstrate that use of a water equivalent cylinder to estimate the dose of an object would yield the correct dose to that object. This question was evaluated by Wang et al., who used Monte Carlo and analytical approaches to demonstrate that because clinical CT systems use a beam-shaping filter, use of water equivalent diameter to estimate an object's dose indeed yields an accurate value for absorbed dose [9].

2.1. Determination of D_w from the CT Image

The attenuation values, or CT numbers, in the axial CT image are expressed using a special unit known as Hounsfield Units (HU):

$$CT\left(x,y\right) = \left(\frac{\mu\left(x,y\right) - \mu_{water}}{\mu_{water}}\right) \times 1000 \tag{1}$$

where $\mu(x,y)$ is the linear attenuation coefficient for a voxel in an axial CT image at position (x,y). Because $\mu(x,y)$ is normalized to the attenuation of water in the definition of CT number, A_w can be represented in terms of CT numbers, as shown in Equation 2,

$$A_w = \sum \left[\frac{\mu\left(x,y\right)}{\mu_{water}}\right]^{\alpha} \times A_{pixel} \tag{2a}$$

$$= \sum \left(\frac{CT\left(x,y\right)}{1000} + 1\right)^{\alpha} \times A_{pixel} \tag{2b}$$

where A_{pixel} is the area of a pixel in the CT image and CT(x,y) is the CT number of a voxel. The parameter α determines the weighting of the linear attenuation coefficients relative to water. In previous works on the estimation of A_w from CT images [5–7], a linear dependence ($\alpha = 1$) was assumed. Wang et al. used analytical and Monte Carlo methods to demonstrate that this assumption is valid when a bowtie filter is used for CT scanning [9].

A_w can be calculated using the mean CT number within a region of interest (ROI). The ROI must be large enough to include the entire patient cross section, but should not include irrelevant objects such as the patient table, since it is only the dose to the patient that is of interest. Equation 2b can then be expanded as,

$$A_w = \sum \left(\frac{CT\left(x, y\right)}{1000} + 1 \right) \times A_{pixel} \quad \text{(3a)}$$

$$= \frac{\sum CT\left(x, y\right)}{1000} \times A_{pixel} + \sum A_{pixel} \quad \text{(3b)}$$

$$= \frac{1}{1000} \times \frac{\sum CT\left(x, y\right)}{N_{pixel}} \times \left(N_{pixel} \times A_{pixel} \right) + \left(N_{pixel} \times A_{pixel} \right) \quad \text{(3c)}$$

$$= \frac{1}{1000} \overline{CT\left(x, y\right)}_{ROI} \quad A_{ROI} + A_{ROI} \quad \text{(3d)}$$

where $\overline{CT\left(x, y\right)}_{ROI}$ is the mean CT number in the ROI, N_{pixel} is the number of pixels in the region of interest, and $A_{ROI} = \sum A_{pixel}$, which is the total area of the ROI. The ROI may include the air surrounding the patient, since voxels that have an attenuation coefficient of nearly zero negligibly change the value of the sum in Equation 2b. The water equivalent diameter, D_w is equal to:

$$D_w = 2 \sqrt{A_w / \pi} \quad \text{(4a)}$$

$$= 2 \sqrt{\left[\frac{1}{1000} \overline{CT\left(x, y\right)}_{ROI} + 1 \right] \frac{A_{ROI}}{\pi}} \quad \text{(4b)}$$

Thus, D_w of an object can be calculated from the mean CT number in an ROI containing that object. The mean CT number can be evaluated using tools readily available on most CT

operator consoles or workstations, although this would require user manual intervention, which may not be practical. Alternatively, automatic segmentation algorithms could be used.

2.2. Determination of D_w from the CT Localizer Radiograph

Estimating D_w from axial CT images can only be performed after the CT scan is finished and images are reconstructed. However, knowledge of patient attenuation may be desirable in order to calculate SSDE prior to scanning. Two studies have shown that patient attenuation can be estimated using the CT localizer radiograph, although the results were not described in terms of D_w [6,7]. Wang et al. derived the expected value of D_w from the CT localizer radiograph image and demonstrated agreement with the results obtained from the axial CT image using a range of phantoms [8, 9].

Conceptually, each horizontal line of a CT localizer radiograph image is associated with a different section of anatomy along the z-axis of the patient; the section width along the z-axis is determined by the collimation settings used by the manufacturer. Each pixel in a CT localizer radiograph image is associated with an x-ray path from the focal spot, through the patient, to the detector, and the pixel value is proportional to the total attenuation along this x-ray path (typically referred to as a ray sum). The width of each ray sum is determined by the detector spacing S. The total attenuation through a patient is the sum of the linear attenuation coefficients (μ_i) of each tissue type times their associated lengths (l_i) along the path of the x-ray beam. It should be noted that the line integral of the attenuation in a CT localizer radiograph will include attenuation of the patient table that must be accounted for in the calculation of D_w.

Because A_w is linearly dependent on the object attenuation, the equivalent path length through water can be found by normalizing the total attenuation by μ_{water}, which is represented by the water equivalent length (L_w),

$$L_w = \int \mu_i \, dl / \mu_{water} \quad (5)$$

Assuming that the pixel values in an appropriately scaled CT localizer radiograph represent ray sums having a mean width at isocenter equal to the detector spacing S at isocenter, the water equivalent area associated with each ray sample is equal to $L_w \times S$. An estimate of the overall water equivalent attenuation of the patient at a particular position along the z-axis can be determined by summing the attenuation associated with each ray sample. That is,

$$A_w = \sum L_w \times S \quad (6)$$

where the sum is taken over a horizontal line of a CT localizer radiograph. Having computed A_w, D_w can be easily obtained using Equation 4a.

3. Comparison between Geometric- and Attenuation-based Metrics

Using Monte Carlo simulation, the average dose to simulated abdomen and thorax phantoms was calculated for adult and pediatric models, as described in Khatonabadi et al. [10]. A bowtie filter was used for all abdomen and thorax phantoms in the Monte Carlo method to simulate a realistic scanning scenario. The D_W of each virtual phantom was calculated with Equation 4b. The radiation dose to a water cylinder was obtained for each D_W value using the same Monte Carlo method and compared with the dose to its corresponding abdomen or thorax phantom. The use of attenuation information did not affect the accuracy of dose estimates in the abdomen, but, as expected, it increased the accuracy of dose estimates for the thorax.

In separate work using patient image data and a validated Monte Carlo technique for estimation of organ dose, Khatonabadi et al. [10] calculated dose to the lungs and kidney for both fixed tube current and modulated tube current scenarios. The correlation of two patient size metrics (effective diameter derived only from cross-sectional area and D_W derived using Equation 4b) to normalized organ dose was determined using the mean $CTDI_{vol}$ from the thoracic (lung) or abdominal (kidney) regions. For the kidney, better correlation was observed for D_W than for effective diameter ($R^2 = 0.89$ and 0.78, respectively). For the lung, better correlation was again observed for D_W compared to effective diameter ($R^2 = 0.92$ and 0.78, respectively).

These and other Monte Carlo studies are consistent with theoretical predictions that dose is most accurately estimated when patient attenuation is taken into account in estimating patient size.

4. Comparison between D_W from CT Localizer Radiographs and D_W from CT Images

Wang et al. studied three sets of abdomen phantoms and two types of thorax phantoms for size ranges from neonate to large adult (Figure 1) [8]. The AP and lateral dimensions and cross-sectional area of the abdomen and thorax phantoms were measured from CT images. Calibration of CT localizer radio-graph pixel values to water thickness was performed using a series of custom-made, torso-shaped phantoms filled with water, which were used to create a variety of thicknesses of water along the AP axis (Figure 1b). For each of the phantoms, effective diameter was calculated using two geometry-based methods: (1) taking the average of lateral and AP dimensions of the phantom and (2) calculating from the cross-sectional area of the phantom. Two attenuation-based methods were used to calculate D_W: (1) using the CT image and (2) using the CT localizer radiograph image.

Calculated values of effective diameter and D_W for the abdomen and thorax phantoms are listed in Table 1 (abdomen) and Table 2 (thorax). Results were normalized to D_W values calculated using the CT images. For abdomen phantoms, D_W was consistently larger than the effective diameter, although the difference was no larger than about 5% (Table 1). In thorax phantoms, effective diameter, which considers only geometry, was 4.3% to 21.5% larger than D_W, which takes into account object attenuation (Table 2). For thorax and abdomen

phantoms, D_W calculated using the CT localizer radiograph was very similar to the CT image-based method, which is consistent with the theoretical equivalence of these two methods shown in Section II. D_W calculated using the CT localizer radiograph was consistently larger compared to D_W calculated using the CT image, although the difference was no larger than 4%. Figure 2 provides a visual comparison between effective diameter and D_W in a patient data set: effective diameter is equivalent to D_W in the abdomen, but is greater than D_W in the thorax. Thus if attenuation is not considered, patient size will be overestimated in the thorax, leading to an underestimation of SSDE.

5. Implementation Issues and Recommendations When a CT Localizer Radiograph is Used to Determine Patient Size

5.1. Measurement of AP or Lateral Dimensions from a CT Localizer Radiograph

The American College of Radiology (ACR) has developed a CT Dose Index Registry for the purposes of gathering large samples of CT dose index data from across the nation. These data will greatly facilitate the establishment of a national reference and achievable dose levels [11–13]. However, there are currently no established measures of patient size or surrogates, such as weight or body mass index, recorded in the DICOM metadata. This presents a considerable barrier to the use of the acquired data, as binning of similarly sized patients cannot be accomplished without prohibitively large amounts of manual data entry to inform the registry of each patient's size. Thus, in the absence of automated methods of calculating D_W being widely available, the data collection software was modified by the ACR to allow the collection of localizer radiographs (starting with Triad version 4.0). Although CT localizer radiographs are not required for the registry, they are highly encouraged, as this allows more meaningful analyses of registry data according to patient size categories, or by calculating SSDE.

To facilitate collection of AP or lateral dimensions from a transmitted lateral or AP CT localizer radiograph, respectively, the registry software uses a segmentation algorithm to automatically extract a patient's body contour and measure a patient's width. Christianson et al. evaluated a range of segmentation techniques for robustness and accuracy across a range of patient data sets [14]. Additionally, they determined an empirical correction factor for use in the thorax to account for the differences in tissue density between water and lungs [14]. A pixel value threshold of 30% of the maximum value along each row of the CT localizer radiograph and a correction factor of 0.85 was used to adjust for the lower attenuation of lung tissue. SSDE values calculated based on their segmentation and correction method agreed with attenuation-based methods of estimating D_W from cross-sectional CT images to within 5% [14].

5.2. Measurement of D_W from the CT Localizer Radiograph

Prior to the scan, some estimate of patient attenuation may be desired, either for determining the scanner output that should be delivered to achieve the required level of image quality or for calculating SSDE prior to scan initiation. The CT localizer radiographs are created from patient attenuation measurements and are used by manufacturers to prescribe the tube

current modulation to be used in the patient scan [15]. Thus, they can be used to estimate D_w. Several important issues, however, may limit the accuracy of this approach.

5.2.1. Need for Calibration of CT Localizer Radiograph Pixel Values in Terms of Water Attenuation—Unlike CT numbers, the pixel values of a CT localizer radiograph image are not reported in absolute units; the relationship between attenuation and pixel value can vary among scanner models and manufacturers. Therefore, the CT localizer radiograph-based method to estimate D_w is not recommended for the user community because it requires careful calibration of CT localizer radiograph pixel values in terms of water attenuation. Although this can be established through a series of experiments, the constancy of such calibrations over time, across software versions, or even across scanners from the same manufacturer, has not been established. Therefore, we recommend that the CT localizer radio-graph image method to estimate D_w be implemented only by scanner manufacturers, who know the relationship between water attenuation and CT localizer radiograph pixel values.

5.2.2. Need to Account for Patient Table Attenuation—The pixel values in the CT localizer radiograph are a measure of the total x-ray attenuation of all material located between the x-ray source and the detector. This includes the attenuation of the patient and the patient table supporting the patient. Thus, in order to estimate only the attenuation of the patient, the attenuation of the patient table must be known for each z position of the table, expressed in terms of water equivalent path length and subtracted from the total attenuation measured in the CT localizer radiograph. Failure to correct for the attenuation of the table will cause greater errors in estimating D_w for smaller patients than for larger patients.

5.2.3. Need for Use of Non-edge-enhanced CT Localizer Radiographs—To assist the operator in identifying anatomic landmarks, various edge-enhancement filters are applied by manufacturers to CT localizer radiograph attenuation measurements. These filters can modify the relationship between attenuation and pixel value in a nonlinear fashion. Hence, in order to estimate patient attenuation from a CT localizer radiograph image according to equations 5 and 6, non-edge-enhancing image filters must be used. However, non-edge-enhanced CT localizer radiograph images are not currently available from all manufacturers.

5.2.4. Need for Careful Patient Centering or for Acquiring Orthogonal CT Localizer Radiographs—Clinically, the patient table is not always positioned such that the patient is centered in the gantry (Figure 3) [7]. Moving a 30-cm diameter water cylinder 5 cm and 10 cm closer to the x-ray tube was shown to lead to overestimation of D_w by 4.5% and 9.9%, respectively [8]. Moving the table 5 cm farther away from the tube led to underestimation of D_w by 4.2% [8]. These errors happen because CT systems calibrate S (detector spacing) with respect to isocenter. When the patient is farther away from the tube relative to the distance from the tube to isocenter (Figure 3a), the patient's image is minified, and the CT localizer radiograph underestimates the amount of attenuation within the field of view (FOV). Conversely, when the patient is closer to the tube compared to the distance to isocenter (Figure 3c), the patient's shadow is magnified, and the CT localizer radiograph overestimates the amount of attenuation in the FOV. Thus, even if the CT localizer

radiograph pixel values are properly calibrated in terms of water attenuation, the magnification error caused by patient mis-centering could lead to errors in size estimation [7, 16].

Since it is difficult to guarantee patient centering in practice, correction for this potential error is important. Li et al. proposed a correction algorithm which requires the use of an orthogonal CT localizer radiograph to estimate and correct for patient mis-centering [16]. They found that the error in D_W was linearly dependent on the table height offset caused by mis-centering, but that the use of an orthogonal CT localizer radiograph to more accurately determine the width of each ray sum reduced this error. Because patients are typically easier to center on the table in the left to right direction compared to in the anterior to posterior direction, a lateral CT localizer radiograph may be less susceptible to mis-centering errors.

5.2.5 Recommendation Regarding Use of CT Localizer Radiograph Images to Estimate D$_W$—This task group considers the value of D_W determined from full FOV axial CT images to be the reference standard for measuring patient x-ray attenuation.

This task group also recognizes that the CT localizer radiograph can be used to calculate reasonable estimates of D_W, but that potential errors can occur when using a CT localizer radiograph to estimate D_W. Because there are some advantages in using CT localizer radiograph images to estimate D_W (e.g., estimates of SSDE can be provided prior to the scan acquisition), and since most manufacturers already use the CT localizer radiograph to estimate attenuation for use with automated exposure control systems, the task group considers this to be an acceptable approach, provided that the implemented method can be shown to determine D_W to within 20% of the reference value over a range of patient sizes and centering conditions. The task group strongly recommends that potential sources of error be taken into account by:

a. calibrating CT localizer radiograph pixel values in terms of water attenuation,

b. accounting for patient table attenuation in the calculation of D_W,

c. using a non-edge-enhancing filter to produce the image used to estimate D_W, and

d. mitigating patient centering issues through acquiring two orthogonal radiographs or implementing an appropriate correction technique.

Estimates of D_W made from CT localizer radiograph images prior to the scan should be replaced after the scan, when possible, provided that D_W is computed using a full field-of-view (FOV) reconstructed axial image or projection data (which will be discussed in the next section).

6. Implementation Issues and Recommendations when a CT Acquisition is Used to Determine D_w

6.1. Use of Acquisition Data in Projection Space or Image Space

Fundamentally, all information related to the patient's size and attenuation is contained in both the projection and image domains. When care is taken (1) to avoid truncation of views or anatomy (e.g., by making sure that the patient's anatomy is fully contained in the projection data set and in the reconstruction FOV) and (2) to ensure accurate attenuation calibrations (e.g., in water and air) and corrections (e.g., scatter and beam hardening) are performed, the measurements of attenuation should be similarly accurate, whether performed using a reconstructed CT image or from the projection data used to reconstruct the CT image. This assumes that the reconstruction process, including the reconstruction kernel applied, introduces no artifacts or other errors in CT number accuracy. This is typically a safe assumption; however, some reconstruction kernels are edge-enhancing or otherwise nonlinear and can alter CT number accuracy, which is not acceptable for the task of using CT numbers to estimate patient attenuation.

Thus, if images are to be used to estimate D_w, the FOV should cover all patient anatomy in the beam, and a linear, quantitative kernel must be used, possibly requiring additional reconstruction data sets. For the user community, such axial CT images provide the easiest route to determining D_W off-line, using either a user-developed software tool or a third-party software package. In this case, extra reconstructions may be necessary and must be integrated into an acceptable clinical workflow.

For manufacturers, the most efficient commercial implementation may be to perform the calculations in projection space, thus avoiding the need to reconstruct any additional (i.e., full field of view or different kernel) images. In the remaining sections of this report, measurement of D_w from a reconstructed axial CT image is described in detail. It should be understood that sinogram and image data are complementary forms of the same data set, and that either form could be used, provided that the described criteria (such as not truncating any patient information and not including attenuation information for objects that are not part of the patient, like the patient table) are able to be met.

6.2. Measurement of D_w from a Reconstructed Axial CT Image

6.2.1. Need for a Full Field of View (FOV) Reconstruction—Use of the reconstructed axial CT image is an accurate way to estimate the patient attenuation because, by definition, an axial CT image gives a cross-sectional map of tissue attenuation. In order to provide an accurate estimate of D_w, it is essential that all patient tissue be included in the reconstructed image. This requires that a full FOV image encompassing the entire patient cross section be reconstructed. For clinical applications where a full FOV image is not reconstructed, such as for imaging of the heart or spine, an additional reconstruction using the full FOV would be required. To minimize additional reconstruction time, an image thickness of not greater than 5 mm and a nonoverlapping reconstruction interval are recommended. Off-line calculations of D_w can then be performed using these full FOV images.

It is anticipated that D_W will eventually be automatically calculated by the scanner for each reconstructed image and the values stored in yet-to-be-specified DICOM data elements. In this scenario, the manufacturer could automatically reconstruct a full FOV image series using a wide image thickness, nonoverlapping interval, and possibly even a decreased matrix size (e.g., 256×256) for the sole purpose of calculating D_W. This process could be fully automated in such a fashion that the user would not ever see this special "low resolution, full FOV image series." This is analogous to the current practice by some manufacturers of reconstructing very thin axial images in a hidden series from which to generate coronal or sagittal images.

As noted in Section 6.1, the manufacturer could calculate the total attenuation in a projection that includes all of the patient (and the table) and then subtract the known attenuation level of the table off of the integrated attenuation measurement. Details with regard to the numbers of projections required, the preferred projection orientation(s), geometrical factors (e.g., fan or rebinned parallel data sets) are left to the manufacturers' discretion, albeit no matter which method is selected, the final implementation must yield D_W results that agree with those calculated using full FOV images to within 10%.

6.2.2. Need to Omit Non-patient Sources of Attenuation from the Measurement
—When using the axial CT image to estimate patient attenuation, care should be taken to define the ROI to include all of the patient, yet minimal amounts of other attenuating materials, such as the patient table. Including large amounts of the patient table can lead to overestimation of patient attenuation, particularly for very small pediatric patients due to their relatively small size. Li et. al. [16] showed that the CT table can contribute up to 12% of the total attenuation for small objects. A manually drawn ROI that conforms more closely to the patient contour than a circular ROI could be used, but this requires more time and effort by the user. As shown in Wang et al., placing a circular ROI around the patient, without attempting to exclude all portions of the patient table, provides acceptable results [9].

Alternatively, automated methods can perform calculations on a voxel-by-voxel basis, as opposed to having to measure the ROI mean and area, as required by Equation 4b. Instead, voxel-by-voxel calculations can be performed using Equation 3a: In this manner, no ROI needs to be drawn manually by a user or automatically through boundary recognition. Voxels having CT numbers that are essentially air can be identified and ignored. Thus, a simple thresholding can remove all voxels with a CT number below, for example, −900 HU. Then Equation 3d can be used with the remaining voxels in the image. As discussed above, it improves accuracy if the table is not included for very small patient sizes, and table removal could be implemented through manufacturer-provided tools or a table recognition algorithm in the software.

The presence of foreign objects in the beam, particularly highly attenuating objects such as metallic implants or bismuth shields, pose a fundamental limitation to the accuracy of D_W and SSDE calculations. Methods to address these limitations are not addressed by this task group report. However, it should be noted that such overestimation of D_W will lead to an underestimation of SSDE. Conversely, in very large patients having tissue outside the scan

FOV (typically 50 cm in diameter), the total attenuation in the beam may be underestimated, which could lead to a small overestimation of SSDE.

7. Longitudinal Position at Which to Measure D$_w$

Patient dimension and attenuation can vary considerably along the longitudinal axis. When tube current modulation is used, scanner output also varies along the longitudinal axis of a patient according to changes in patient attenuation.

The mean scanner output per rotation normalized to the tube-current time product (CTDI$_{vol}$/mAs) is easily measured or can be determined from the scanner's accompanying documents. The mean effective mAs per rotation is easily calculated by the scanner, or it can be determined from image header information. By multiplying the mean effective mAs per rotation at every image location along the z-axis with the nominal CTDI$_{vol}$/mAs, the scanner output at each location can be calculated. This information can be recorded as a series of CTDI$_{vol}$ values for each reconstructed image, which we refer to as CTDI$_{vol}$(z). SSDE can then be calculated at each position z along the longitudinal direction,

$$SSDE(z) = f_{D_w}(z) \cdot CTDI_{vol}(z) \quad (7)$$

where f_{D_w}(z) is the size-specific conversion factor from AAPM Report 204. The mean SSDE over the entire scan range can be expressed as

$$\overline{SSDE} = \frac{\sum_{z=1}^{N} SSDE(z)}{N} \quad (8)$$

where N is the total number of images.

This approach requires that D_W be calculated for each table location. This would allow the determination of the mean SSDE over the scan range using the $f_{D_W}(z)$ and $CTDI_{vol}(z)$ at each longitudinal position.

Leng et al. showed, however, that the mean SSDE over a scan range calculated using Equation 8 correlated extremely well with a mean SSDE calculated using the D_W from a central image in the scan range and the mean CTDI$_{vol}$(z) over the entire scan range [17]. Thus, for the purpose of estimating a single SSDE value for a given patient, the "shortcut" of using the scanner-reported mean CTDI$_{vol}$ and the D_W from a central location in the scan range (e.g., the center of the prescribed scan range on the CT localizer radiograph or the central image of the CT images) appears acceptable. It is highly recommended, however, that z-specific values of D_W be provided in a standardized DICOM tag. This may provide future opportunities to estimate organ doses by having attenuation (and mAs information) along the longitudinal patient axis [18].

8. Summary and Road Map for Adoption

This task group was charged with developing and analyzing robust and scientifically sound metrics for automatically estimating patient size in CT that would account for patient attenuation and allow routine determination of SSDE with little or no human intervention (e.g., no drawing of ROIs and no manual calculations). Specifically, the size metric *water equivalent diameter* was described and two approaches for obtaining this value were evaluated: (a) one based on the CT localizer radiograph and (b) one based on an axial CT image reconstructed using a full FOV.

Pragmatically, patient attenuation needs to be determined using automated methods implemented on the scanner so that a patient-specific size metric can be included as part of the dose report information. This would allow SSDE to be automatically calculated and saved in DICOM data elements. A standardized methodology should be approved by an appropriate standards organization and implemented in a consistent manner by all scanner manufacturers.

This task group recommends:

1.   Adopt water equivalent diameter as the preferred patient size metric. This value reflects the x-ray attenuation of the patient and is a sound descriptor of patient size.

2.   The preferred method to obtain water equivalent diameter values would be that they be calculated from projection or full FOV CT image data. This approach would provide a more robust estimate of patient size compared to use of the CT localizer radiograph. This task group understands that there may be practical limitations to implementing this approach, including the possible need to perform additional full FOV images, which would increase reconstruction time and create more images to review, transfer, and archive.

3.   A reasonable method to obtain estimates of water equivalent diameter using the CT localizer radiograph must take into account potential sources of error by requiring the following:

   a.   Pixel values are calibrated in terms of water attenuation.

   b.   The attenuation of the table is accounted for in the calculation of D_w.

   c.   Non-edge-enhancing filters are used in producing the image used to estimate D_w.

   d.   Patient centering issues are mitigated by acquiring two orthogonal radiographs or using some other method of correcting for magnification error.

4.   If the CT localizer radiograph method is used, then the D_w and SSDE values after the scan should be automatically calculated and used to

replace pre-scan values for D_W and SSDE (estimated using the CT localizer radiograph); these post-scan results are the ones that should be stored in DICOM data elements.

5. If the difference between the pre- and post-scan values for D_W and SSDE can be shown by a manufacturer to be consistently less than 10% of the pre-scan values across a range of patient size and habitus, the pre-scan values may be used as the final values.

6. The preferred approach is to obtain D_W values at multiple locations along the z-axis of the patient, at intervals not to exceed 5 mm; however, reasonable estimates of SSDE may be obtained with a single D_W value from the central image of the scanned range.

Both scanner manufacturers and third-party suppliers of dose metric tracking tools need to quickly develop an implementation strategy to allow the appropriate DICOM working groups to include the needed data fields in the next version of the DICOM standard.

To facilitate testing of the various implementation strategies, deidentified test data and the associated reference D_W values should be made available to manufacturers for use in their development and testing processes. Systems that meet the criteria described in this document and produce root mean square errors of less that 10% relative to the reference values for the test patient cases are considered to be compliant with this report's recommendations.

Considering the rapid rate of adoption of SSDE in the clinical and research communities, and given the continued interest in assessing patient-specific dose models, it is imperative that scanner manufacturers and other dose metric tracking software providers automatically calculate D_W at z-axis intervals of no larger than 5 mm and store these data in clearly labeled DICOM data fields, in both the image and the Radiation Dose Structured Report, as soon as possible. Additionally, scanner manufacturers are urged to provide the angular and longitudinal modulation tube current values in a separate DICOM data array to allow $CTDI_{vol}$ and SSDE to be calculated for z-axis ranges that are clinically relevant, e.g., over the entire thorax or over just the heart. While the mean tube current value per image, or over a 5 mm z-axis interval, provide important and useful information, the individual tube current values for different view angles (x-ray source angles) should be provided at reasonable intervals (such as every 1 degree or every 5 degrees).

The strongly expressed goal of the CT stakeholder community is that this work be completed and begin shipping on software versions released in 2016. This will provide users sophisticated metrics for assessing size-specific patient doses in an automated and reproducible manner for use in many quality and safety initiatives.

Appendix

Clarification Regarding the Use of the Term "Effective Diameter" Instead of D_W in AAPM Report 204

The conversion factors in AAPM Report 204 were derived with water- or tissue-equivalent materials. In the case of the PMMA phantoms used by Toth and Strauss, the attenuation was expressed in terms of water attenuation by correcting for the increased density of PMMA relative to water. Thus, although the phrase "effective diameter" was used in the report, all measurements and calculations were in terms of D_W. Thus, in **Equation A–1** of AAPM Report 204, which describes the best fit to the data shown in Figures 4–6, patient size is already expressed in terms of D_W:

$$y = a \times e^{-bx} \quad \text{(A-1)}$$

where the coefficients are given in the table below:

Figure Number	X parameter	Y parameter	a	b
4	D_W (cm)	Conversion Factor	3.704369	0.03671937
5	D_W (cm)	Conversion Factor	4.378094	0.04331124
6	D_W (cm)	Conversion Factor	1.874799	0.03871313

When D_W values are available, they should be used in Tables 1 and 2 and in Equation A–1 of AAPM Report 204. No additional corrections are required when substituting D_W for effective diameter.

In the thorax, use of effective diameter instead of D_W will lead to an overestimation of patient attenuation and an underestimation of SSDE. However, as noted in AAPM Report 204 and shown in Table 2 of this report, when attenuation is not considered, errors in the estimate of thoracic size are below 20%. In the abdomen, the errors are much smaller, in the range of a few percent. Thus, even though use of D_W is recommended, when only geometric data are available to the user, it is still reasonable to calculate SSDE based on any of the geometric input parameters shown in Report 204: AP, LAT or AP+LAT, or effective diameter.

Additionally, when the cross-sectional area of the patient is known (without attenuation information), use of effective diameter as the input parameter is preferable to the use of AP, LAT, or AP+LAT. This is because knowledge of cross-sectional area allows calculation of effective diameter directly, without any assumptions regarding the shape of the patient (i.e., without assuming that the patient cross-section is elliptical).

In all cases, knowledge of both patient attenuation and cross-sectional area allows the most accurate calculation of D_W and hence the most accurate calculation of SSDE.

Table A-1

This table provides conversion factors based on the use of the 32 cm diameter PMMA phantom for CTDI$_{vol}$. Other than the clarification that the effective diameter in the data table is equivalent to D_w, it is identical to Table 1 in AAPM Report 204.

Table 1A			Table 1B			Table 1C			Table 1D	
Lat + AP Dim (cm)	Dw (cm)	Conversion Factor	Lateral Dim (cm)	Dw (cm)	Conversion Factor	AP Dim (cm)	Dw (cm)	Conversion Factor	Dw (cm)	Conversion Factor
16	7.7	2.79	8	9.2	2.65	8	8.8	2.68	8	2.76
18	8.7	2.69	9	9.7	2.60	9	10.2	2.55	9	2.66
20	9.7	2.59	10	10.2	2.55	10	11.6	2.42	10	2.57
22	10.7	2.50	11	10.7	2.50	11	13.0	2.30	11	2.47
24	11.7	2.41	12	11.3	2.45	12	14.4	2.18	12	2.38
26	12.7	2.32	13	11.8	2.40	13	15.7	2.08	13	2.30
28	13.7	2.24	14	12.4	2.35	14	17.0	1.98	14	2.22
30	14.7	2.16	15	13.1	2.29	15	18.3	1.89	15	2.14
32	15.7	2.08	16	13.7	2.24	16	19.6	1.81	16	2.06
34	16.7	2.01	17	14.3	2.19	17	20.8	1.73	17	1.98
36	17.6	1.94	18	15.0	2.13	18	22.0	1.65	18	1.91
38	18.6	1.87	19	15.7	2.08	19	23.2	1.58	19	1.84
40	19.6	1.80	20	16.4	2.03	20	24.3	1.52	20	1.78
42	20.6	1.74	21	17.2	1.97	21	25.5	1.45	21	1.71
44	21.6	1.67	22	17.9	1.92	22	26.6	1.40	22	1.65
46	22.6	1.62	23	18.7	1.86	23	27.6	1.34	23	1.59
48	23.6	1.56	24	19.5	1.81	24	28.7	1.29	24	1.53
50	24.6	1.50	25	20.3	1.76	25	29.7	1.25	25	1.48
52	25.6	1.45	26	21.1	1.70	26	30.7	1.20	26	1.43
54	26.6	1.40	27	22.0	1.65	27	31.6	1.16	27	1.37
56	27.6	1.35	28	22.9	1.60	28	32.6	1.12	28	1.32
58	28.6	1.30	29	23.8	1.55	29	33.5	1.08	29	1.28
60	29.6	1.25	30	24.7	1.50	30	34.4	1.05	30	1.23
62	30.5	1.21	31	25.6	1.45	31	35.2	1.02	31	1.19

Table 1A

Lat + AP Dim (cm)	Dw (cm)	Conversion Factor
64	31.5	1.16
66	32.5	1.12
68	33.5	1.08
70	34.5	1.04
72	35.5	1.01
74	36.5	0.97
76	37.5	0.94
78	38.5	0.90
80	39.5	0.87
82	40.5	0.84
84	41.5	0.81
86	42.4	0.78
88	43.4	0.75
90	44.4	0.72

Table 1B

Lateral Dim (cm)	Dw (cm)	Conversion Factor
32	26.6	1.40
33	27.6	1.35
34	28.6	1.30
35	29.6	1.25
36	30.6	1.20
37	31.7	1.16
38	32.7	1.11
39	33.8	1.07
40	34.9	1.03
41	36.1	0.98
42	37.2	0.94
43	38.4	0.90
44	39.6	0.87
45	40.8	0.83

Table 1C

AP Dim (cm)	Dw (cm)	Conversion Factor
32	36.0	0.99
33	36.8	0.96
34	37.6	0.93
35	38.4	0.91
36	39.1	0.88
37	39.8	0.86
38	40.4	0.84
39	41.1	0.82
40	41.7	0.80
41	42.3	0.78
42	42.8	0.77
43	43.4	0.75
44	43.9	0.74
45	44.4	0.73

Table 1D

Dw (cm)	Conversion Factor
32	1.14
33	1.10
34	1.06
35	1.02
36	0.99
37	0.95
38	0.92
39	0.88
40	0.85
41	0.82
42	0.79
43	0.76
44	0.74
45	0.71

Table A-2

This table provides conversion factors based on the use of the 16 cm diameter PMMA phantom for CTDI$_{vol}$. Other than the clarification that the effective diameter in the data table is equivalent to D_w, it is identical to Table 2 in AAPM Report 204.

Table 2A			Table 2B			Table 2C			Table 2D	
Lat + AP Dim (cm)	Dw (cm)	Conversion Factor	Lateral Dim (cm)	Dw (cm)	Conversion Factor	AP Dim (cm)	Dw (cm)	Conversion Factor	Dw (cm)	Converion Factor
12	5.7	1.50	6	8.2	1.36	6	5.8	1.50	6	1.49
13	6.2	1.47	7	8.7	1.34	7	7.3	1.41	7	1.43
14	6.7	1.44	8	9.2	1.32	8	8.8	1.33	8	1.38
15	7.2	1.42	9	9.7	1.29	9	10.2	1.26	9	1.32
16	7.7	1.39	10	10.2	1.26	10	11.6	1.19	10	1.27
17	8.2	1.36	11	10.7	1.24	11	13.0	1.13	11	1.22
18	8.7	1.34	12	11.3	1.21	12	14.4	1.07	12	1.18
19	9.2	1.31	13	11.8	1.19	13	15.7	1.02	13	1.13
20	9.7	1.29	14	12.4	1.16	14	17.0	0.97	14	1.09
21	10.2	1.26	15	13.1	1.13	15	18.3	0.92	15	1.05
22	10.7	1.24	16	13.7	1.10	16	19.6	0.88	16	1.01
23	11.2	1.22	17	14.3	1.08	17	20.8	0.84	17	0.97
24	11.7	1.19	18	15.0	1.05	18	22.0	0.80	18	0.93
25	12.2	1.17	19	15.7	1.02	19	23.2	0.76	19	0.90
26	12.7	1.15	20	16.4	0.99	20	24.3	0.73	20	0.86
27	13.2	1.13	21	17.2	0.96	21	25.5	0.70	21	0.83
28	13.7	1.10	22	17.9	0.94	22	26.6	0.67	22	0.80
29	14.2	1.08	23	18.7	0.91	23	27.6	0.64	23	0.77
30	14.7	1.06	24	19.5	0.88	24	28.7	0.62	24	0.74
31	15.2	1.04	25	20.3	0.85	25	29.7	0.59	25	0.71
32	15.7	1.02	26	21.1	0.83	26	30.7	0.57	26	0.69
33	16.2	1.00	27	22.0	0.80	27	31.6	0.55	27	0.66
34	16.7	0.98	28	22.9	0.77	28	32.6	0.53	28	0.63
35	17.2	0.97	29	23.8	0.75	29	33.5	0.51	29	0.61

| Table 2A | | | Table 2B | | | Table 2C | | | Table 2D | |
Lat + AP Dim (cm)	Dw (cm)	Conversion Factor	Lateral Dim (cm)	Dw (cm)	Conversion Factor	AP Dim (cm)	Dw (cm)	Conversion Factor	Dw (cm)	Conversion Factor
36	17.6	0.95	30	24.7	0.72	30	34.4	0.50	30	0.59
37	18.1	0.93	31	25.6	0.70	31	35.2	0.48	31	0.56
38	18.6	0.91	32	26.6	0.67	32	36.0	0.46	32	0.54
39	19.1	0.89	33	27.6	0.65	33	36.8	0.45	33	0.52
40	19.6	0.88	34	28.6	0.62	34	37.6	0.44	34	0.50
42	20.6	0.84	35	29.6	0.60	35	38.4	0.42	35	0.48
44	21.6	0.81	36	30.6	0.57	36	39.1	0.41	36	0.47
46	22.6	0.78	37	31.7	0.55	37	39.8	0.40	37	0.45
48	23.6	0.75	38	32.7	0.53	38	40.4	0.39	38	0.43
50	24.6	0.72	39	33.8	0.51	39	41.1	0.38	39	0.41
52	25.6	0.70	40	34.9	0.48	40	41.7	0.37	40	0.40
54	26.6	0.67	41	36.1	0.46	41	42.3	0.36	41	0.38
56	27.6	0.64	42	37.2	0.44	42	42.8	0.36	42	0.37
58	28.6	0.62	43	38.4	0.42	43	43.4	0.35	43	0.35
60	29.6	0.60	44	39.6	0.40	44	43.9	0.34	44	0.34
62	30.5	0.57	45	40.8	0.39	45	44.4	0.34	45	0.33
64	31.5	0.55	46	42.1	0.37	46	44.8	0.33	46	0.32
66	32.5	0.53	47	43.3	0.35	47	45.2	0.33	47	0.30
68	33.5	0.51	48	44.6	0.33	48	45.6	0.32	48	0.29
70	34.5	0.49	49	45.9	0.32	49	46.0	0.32	49	0.28
72	35.5	0.47	50	47.2	0.30	50	46.4	0.31	50	0.27
74	36.5	0.46	51	48.5	0.29	51	46.7	0.31	51	0.26
76	37.5	0.44	52	49.9	0.27	52	47.0	0.30	52	0.25
78	38.5	0.42	53	51.3	0.26	53	47.2	0.30	53	0.24
80	39.5	0.41	54	52.7	0.24	54	47.5	0.30	54	0.23
82	40.5	0.39	55	54.1	0.23	55	47.7	0.30	55	0.22

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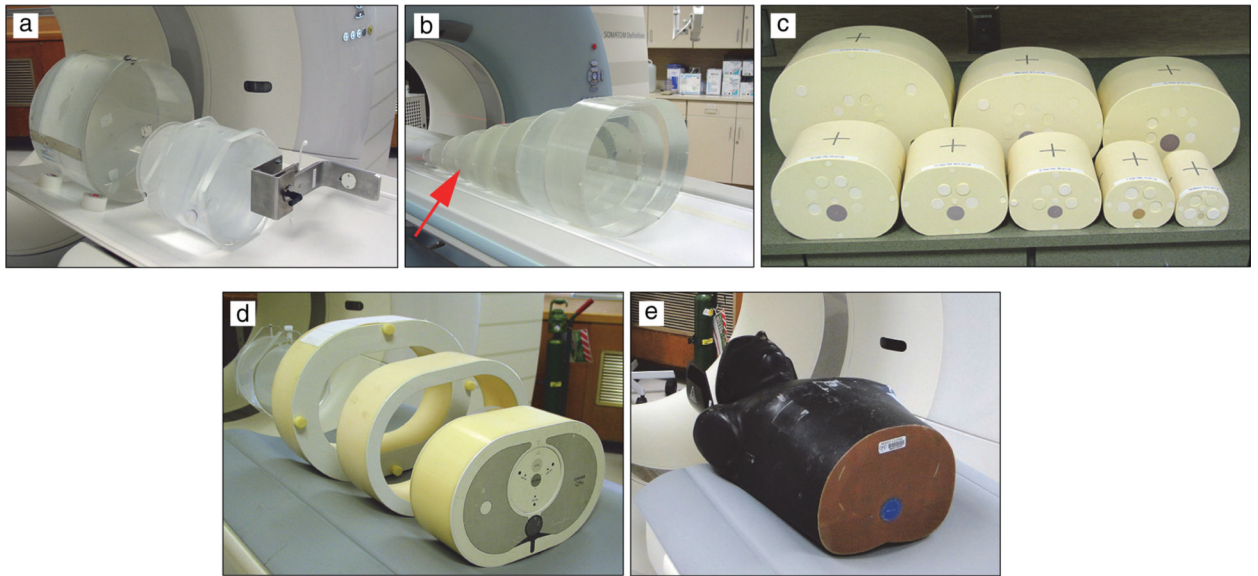
Figure 1.
Phantoms used for estimating D_W from CT localizer radiographs and CT images: (a) water cylinders, (b) torso-shaped phantoms with a PMMA shell filled with water, (c) water equivalent torso phantoms, (d) semi-anthropomorphic thorax phantom with extension rings for mimicking larger patient sizes, and (e) anthropomorphic thorax phantom.
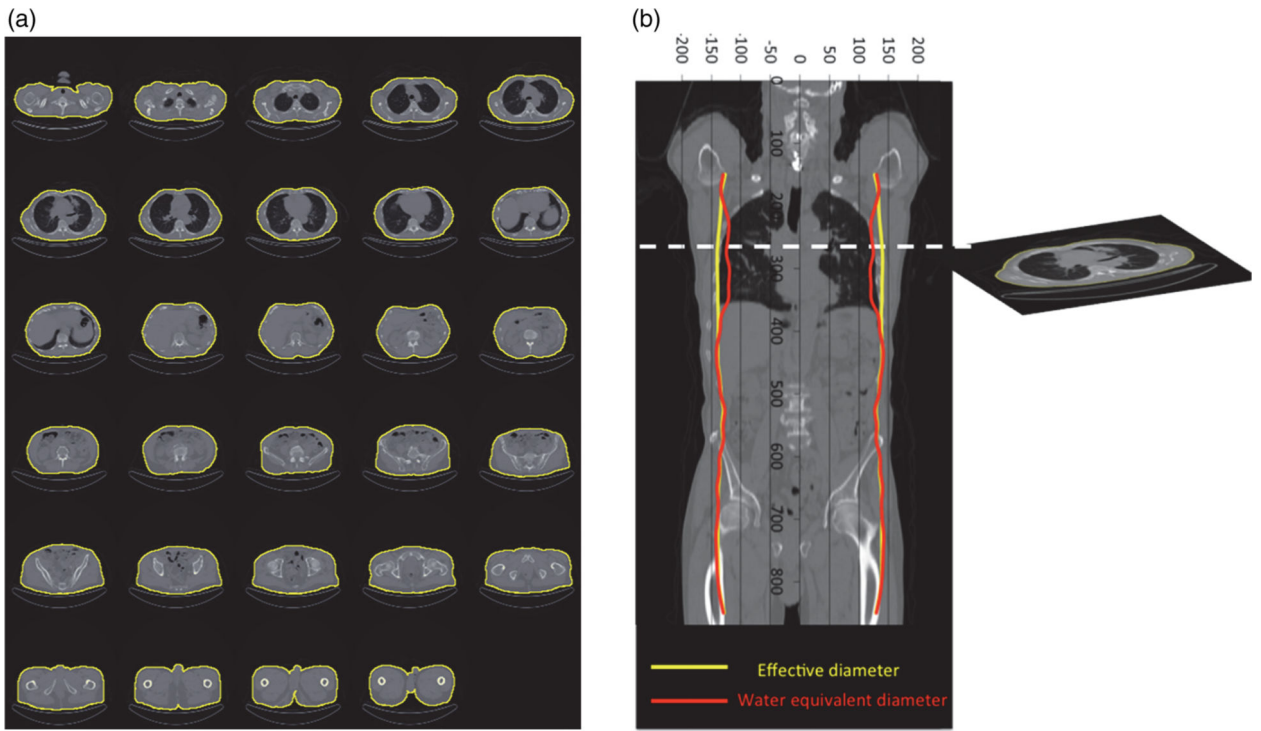
Figure 2.
Comparison of effective diameter and D_W for a patient data set. A threshold of -383 HU was used to remove the patient table for every 10th image in the data set. The segmented ROIs are shown in (a). The area and mean CT number of the ROI in each axial image were used to calculate effective diameter and D_W, which are superimposed on a coronal reformation of the patient data in (b). This visually demonstrates that the use of effective diameter will overestimate D_W in the thorax.
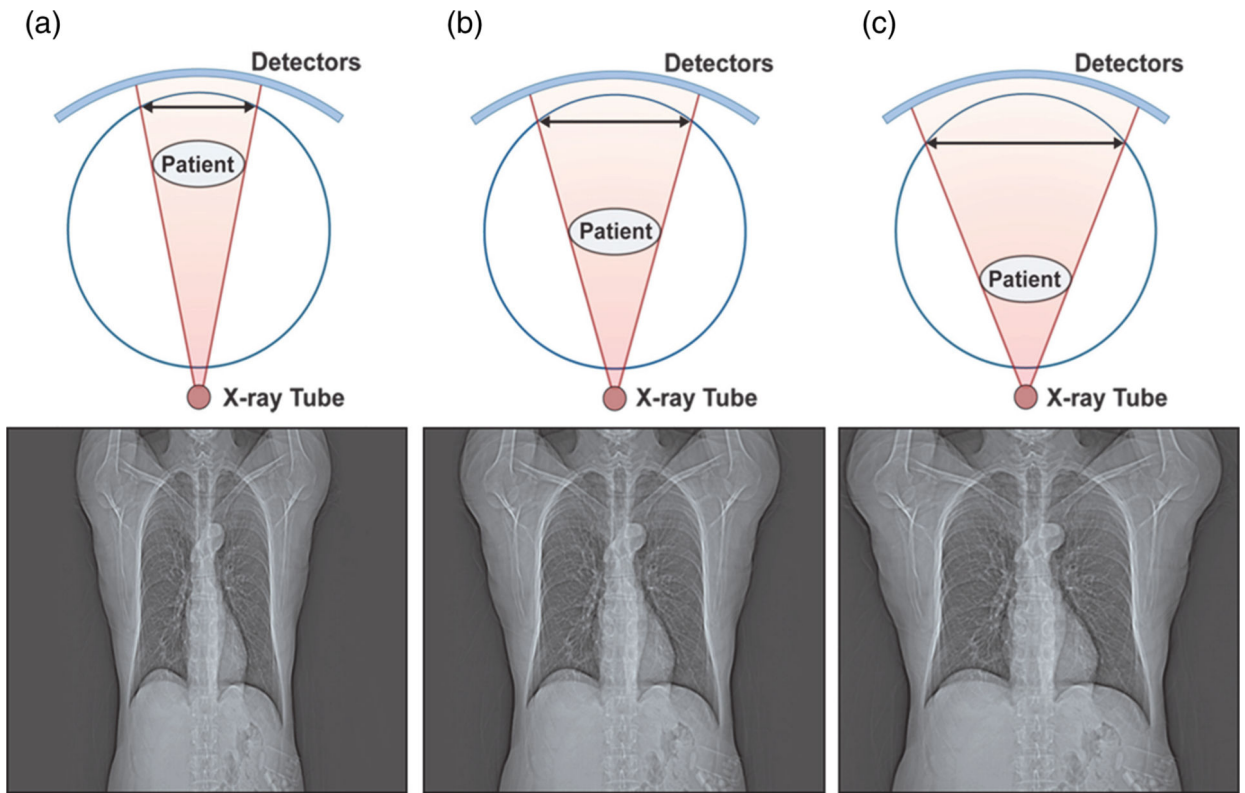
Figure 3.
Minification (a) or magnification (c) of the patient's lateral dimensions occurs when the patient is not well centered in the FOV (b).

Table 1

Abdominal phantom

AP and lateral dimensions, effective diameters calculated using phantom dimensions, and water equivalent diameters (D_W) calculated using CT images and CT localizer radiographs. Effective and water equivalent diameter values were normalized to D_W calculated from the CT images. The absolute value of D_W calculated using the CT image is shown in parentheses. (Adapted with permission from Wang et al. [8].)

Phantom	Phantom Size	Phantom Dimension (cm)		Effective Diameter (%)		Water Equivalent Diameter (%)	
		AP	LAT	(AP + LAT)/2	$\sqrt{AP \times LAT}$	From CT Image**	From CT Localizer Radiograph
Water Cylinder	20 cm	20	20	98.0	98.0	100 (20.4)	101.0
	30 cm	30	30	98.0	98.0	100 (30.6)	102.0
Abdomen Water Phantom	15 cm *^	11.2	15	97.0	98.5	100 (13.5)	103.7
	20 cm *	14.9	20	97.2	98.9	100 (18.0)	102.2
	25 cm *	18.6	25	96.0	98.2	100 (22.7)	100.9
	30 cm *	22.4	30	96.3	98.2	100 (27.2)	100.7
	35 cm *	26.1	35	96.8	98.7	100 (31.6)	101.6
	40 cm *	29.8	40	96.7	98.6	100 (36.1)	101.7
	45 cm *	33.6	45	97.0	99.0	100 (40.5)	102.5
Tissue Equivalent Abdomen Phantom	Newborn	8.9	10.6	96.1	99.0	100 (10.2)	102.9
	1 Yr	11.4	14	95.5	98.5	100 (13.3)	101.5
	5 Yr	13.9	18	95.8	98.8	100 (16.7)	101.2
	10 Yr	15.9	20.6	95.8	98.4	100 (19.1)	101.0
	15 Yr	18.4	24	95.1	98.2	100 (22.3)	100.9
	Small Adult	22	30	95.6	98.5	100 (27.2)	101.1
	Medium Adult	24.9	32.4	96.3	99.3	100 (29.8)	101.7
	Large Adult	30.7	38.9	95.1	98.4	100 (36.6)	101.4
Mean ± Standard Deviation				96.4 ±0.9%	98.5 ±0.4%	-	101.6 ±0.8%

AP = anteroposterior, LAT = lateral

*Lateral width of the phantom

**Absolute value in parenthesis (cm)

^Size of a newborn.

Table 2

Thorax Phantom

AP and lateral dimensions, effective diameters calculated using phantom dimensions, and water equivalent diameters D_W calculated using CT images and CT localizer radiographs. Effective and water equivalent diameter values were normalized to D_W calculated from the CT images. The absolute value of D_W calculated using the CT image is shown in parentheses. (Adapted with permission from Wang et al. [8].)

Phantom	Phantom Size	Phantom Dimension (cm)		Effective Diameter (%)		Water Equivalent Diameter (%)	
		AP	LAT	(AP + LAT)/2	$\sqrt{AP \times LAT}$	From CT Image **	From CT Localizer Radiograph
Tissue Equivalent Thorax Phantom	30 cm *	20	30	119.6	121.5	100 (20.9)	100.0
	35 cm *	25	35	109.9	112.5	100 (27.3)	100.7
	40 cm *	30	40	105.4	107.5	100 (33.2)	101.5
Anthro-pomorphic Phantom		21.6	31.8	104.3	112.1	100 (25.6)	100.0
Mean ± Standard Deviation				109.8 ±7.0%	113.4 ±5.9%	-	100.6 ±0.7%

AP = anteroposterior, LAT = lateral

*
Lateral width of the phantom

**
Absolute value in parenthesis (cm).