Automatic calculation of patient size metrics in computed tomography: What level of computational accuracy do we need?

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Abstract

Objectives: To compare the effectiveness of two different patient size metrics based on water equivalent diameter \(D_w\), the mid-scan water equivalent diameter \(D_{w_c}\), and the mean (average) water equivalent diameter in the imaged region, \(D_{w_{ave}}\), for automatic detection of accidental changes in computed tomography (CT) acquisition protocols.

Methods: Patient biometric data (height and weight) were available from a previous survey for 80 adult chest examinations, and 119 adult single-acquisition chest–abdomen–pelvis (CAP) examinations for two 16 slice scanners (GE LightSpeed and Toshiba Aquillio RXL) equipped with automatic tube current modulation (ATCM). \(D_{w_c}\) and \(D_{w_{ave}}\) were calculated from the archived CT images. Size-specific dose estimates (SSDE) were obtained from volume CT dose index (CTDIvol), using the conversion factors for a patient diameter of \(D_{w_c}\).

Results: CTDIvol and SSDE correlate better with \(D_{w_{ave}}\) than with \(D_{w_c}\). R-squared values of linear fits to CTDIvol of CAP examinations were 0.81–0.89 for \(D_{w_c}\) and 0.93–0.94 for \(D_{w_{ave}}\) (SSDE: 0.69–0.80 for \(D_{w_c}\), 0.87–0.92 for \(D_{w_{ave}}\)). Percentage differences between \(D_{w_c}\) and \(D_{w_{ave}}\) were \(-4 \pm 4\%\) for chest and \(+5 \pm 4\%\) for CAP examinations (in % of \(D_{w_{ave}}\)). However, small \(D_w\) variations translated as larger variations in CTDIvol for these ATCM systems (e.g., a 24% increase in \(D_w\) doubled CTDIvol). The dependence of CTDIvol on \(D_{w_{ave}}\) was similar for chest and CAP examinations performed with similar ATCM parameters, while use of \(D_{w_c}\) resulted in a clear separation of the same data according to examination type. Maximum \(D_w\) variation in the imaged region was 5.6 \pm 1.6 cm for chest and 6.5 \pm 1.4 cm for CAP examinations.

Conclusions: \(D_{w_{ave}}\) is a better metric than \(D_{w_c}\) for binning similar-sized patients in dose comparison studies, despite the additional computational effort required for its calculation. Therefore, when implementing automatic determination of \(D_w\) for SSDE calculations, automatic calculation of \(D_{w_{ave}}\) should be considered.

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1 | INTRODUCTION

Computed tomography (CT) is a powerful diagnostic tool, but CT imaging protocols should be optimized to minimize radiation exposure. Diagnostic reference levels (DRLs) have been a powerful tool in dose optimization, by establishing typical values of volume CT Dose Index (CTDIvol) or dose-length product (DLP), for certain types of examinations performed on standard-sized (70 ± 3 kg) patients.1-4

Modern CT scanners are equipped with automatic tube current modulation (ATCM), which adjusts tube current according to patient size and anatomical region, based on parameters set by the user.5 The functioning of an ATCM system and the conditions to be set depend on the scanner manufacturer. For GE and Toshiba systems, three parameters must be specified: an "image quality index" related to image noise, and the minimum and maximum values of tube current, Imin and Imax. An adequate value of Imin avoids unnecessary dose escalation in large or obese patients. Imin is equally important to prevent excessive noise in smaller patients, particularly in low attenuation regions such as the lungs.6-7 GE defines a parameter called noise index (NI) to specify the noise level, while Toshiba uses the standard deviation (SD).8,9

ATCM systems have some limitations, and pediatric patients need separate imaging protocols, with different parameters (such as lower kV and lower Imin). The range of sizes in pediatric patients is immense, from babies to adolescents, and different protocols should be used according to child size/age.8,10,11 Only adult patients will be considered in the present work.

Optimization of ATCM settings is a time-consuming process involving a radiologist who assesses image quality after each acquisition. This is usually done for only a few patients. If the results are considered acceptable, the protocol is implemented in a provisional fashion. Postacquisition assessment of examination doses and image quality is continued for some time, to confirm that settings are optimized for all patient sizes.7 In this context, it is useful to have reference dose levels for different-sized adult patients. The American Association of Physicists in Medicine (AAPM) lists approximate reference values for different weight ranges in some typical examination protocols.12,13

Reliance on ATCM systems increases the potential detriment of nonoptimized settings and accidental changes to previously optimized protocols, as follows. An accidental increase in target image noise will result in a degradation of image quality for all patients, which should be quickly detected by radiologists. However, a decrease in target image noise will increase examination doses and image quality for smaller adults, while scanner output for large patients is limited by Imax. Likewise, an unnecessarily high value of Imin increases examination doses for small adults, with no degradation of image quality. Both situations lead to saturation of the tube current (at Imax or Imin) for an increased number of patients,7 but this may go unnoticed under a heavy workload, or be dismissed through overconfidence in the automated system.

With the introduction of PACS (Picture Archiving and Communication Systems) in radiology, several vendors have developed radiation dose index monitoring (RDIM) software, which collects dosimetric information from imaging studies and stores it in a relational database.14 RDIM systems are a powerful tool to identify accidental changes and outliers, and determine where optimization is needed. However, patient biometric data (height and weight) are not usually available in PACS. Therefore, an accidental change which affects only small adults is hard to recognize quickly, because individual examination doses are still in the expected range (e.g., a 50-kg adult imaged with a CTDIvol adequate for a 90-kg patient). Naturally, dosimetric data from thousands of examinations will include patients of all sizes and can be compared between different institutions and scanners. But this is population-dependent and also impractical for quick detection of changes and nonoptimized protocols.

The AAPM Task Group 204 proposed the use of size-specific dose estimates (SSDE) for patient dose comparisons. SSDE is an estimate of patient dose at the center of the imaged region, obtained from CTDIvol using conversion factors f(Dvol) related to the effective diameter of the patient, calculated from the measured anteroposterior (AP) and lateral (LAT) patient dimensions, Dvol = √(AP·LAT).15 To improve the calculation of SSDE by taking into account patient attenuation, the AAPM Task Group 220 proposed describing patient size in terms of water equivalent diameter (Dw). TG220 also suggested that Dw could be calculated automatically by the CT scanner for all patients, with no user intervention, and the results stored in the DICOM header of CT images.16 An automatically calculated Dw would allow binning of similar-sized patients in RDIM databases for comparison of examination doses. SSDE values for adults may vary with patient size, depending on the ATCM system.17

Leng et al. have shown that SSDE can be calculated with less than 10% error using the examination CTDIvol and the value of Dw obtained from the mid-scan CT slice (Dw,I).16 However, values of Dw along the imaged region, Dw[z], may be useful for estimating organ doses.16,18

Obtaining Dw[z] values requires longer computational times, but it also allows calculation of the mean value (average) of Dw[z], Dwave. As the response of ATCM systems is based on patient attenuation, Dwave is the quantity more closely related to the examination’s mean CTDIvol.19 Anam et al. recently reported on the implementation of automatic contouring for calculation of Dw, and showed that Dwave could be obtained with reasonable accuracy from only nine images, for head and thorax examinations.20 This is still nine times the computational effort required for calculating Dwave. Differences between Dwave and Dwave were found to be less than 10%,20 which agrees well with data reported by other authors.21
The aim of this study was to compare \( D_{w,c} \) and \( D_{w,ave} \) as patient size metrics, for the purpose of ATCM optimization and detection of accidental changes; and to determine whether the difference between the two is sufficient to justify the additional computational effort required to automatically determine \( D_{w,ave} \) in addition to \( D_{w,c} \). This study also assessed the interdependence of metrics and the feasibility of using \( D_w \) metrics in nonautomated scenarios, for retrospective comparison with older data.

2 | MATERIALS AND METHODS

2.A | Data collection

This study took advantage of existing biometric data, which had been collected during a routine internal survey, after confirmation by the radiologists that image quality was satisfactory. Patient biometric data (height and weight) were available for 80 chest and 119 single-acquisition chest-abdomen-pelvis (CAP) examinations, performed in either of the two scanners available at our institution: a GE Lightspeed in use since 2011 (CT11) and the Toshiba Aquilion RXL acquired in March 2014 (CT14).

The available data, summarized in Tables 1 and 2, pertain only to adult patients (21–89 years, mean 62 years), because pediatric examinations use separate protocols. CAP and chest examinations were chosen because they are frequently performed and because the corresponding protocols are used as a basis for the examination protocols of more complex examinations, which are harder to optimize independently.

Both CT11 and CT14 are 16-slice scanners, equipped with ATCM in both the longitudinal direction (\( z \)-axis modulation) and the perpendicular plane (\( xy \) or angular modulation). The combination of these two is known as 3D modulation. Patients are randomly assigned to one scanner or the other, depending on equipment availability and internal logistics. Two orthogonal scout images (tube positions 0° and 90°) were acquired before each examination, in the order recommended by each manufacturer. The acquisition parameters are summarized in Table 3.

Proper functioning of the ATCM system and scanner indications of dosimetric parameters were checked at acceptance, and then annually, following the protocols and recommendations of the Spanish Medical Physics Society.

To reduce patient dose in CT examinations, it is important to limit anatomical coverage (scan range) to the area of clinical concern. Appropriate restriction of anatomical coverage minimizes the scan length, whereas the optimization of ATCM parameters is reflected in the examination’s mean CTDIvol. Both influence the dose-length product (DLP). In this work, the examination CTDIvol (mean CTDIvol for the 32 cm diameter CTDI phantom) was chosen as the dosimetric parameter of interest and obtained from the dose summary archived in PACS.

2.B | Retrospective data analysis using attenuation metrics

Reconstructed CT images can be used to calculate \( D_w \), provided the reconstruction kernel is linear and quantitative (not edge-enhancing or otherwise nonlinear). The image series obtained with the SOFT (GE) and FC08 (Toshiba) reconstruction kernels were used for this study.

The field of view (FOV) used in chest and CAP examinations usually includes the outer contour of the patient. Visual observation of the CT images confirmed that, for the majority of the examinations, the whole contour of the skin was visible in the entire imaged region, except near the shoulders. Examinations where a large part of the patient’s contour was outside the FOV were excluded from the dataset. These situations were too rare to justify a correction based on air border proportion, as suggested by Ikuta et al.

According to the AAPM Task Group 220, the water equivalent diameter \( (D_w) \) of an object is related to its water equivalent area \( (A_w) \): \( D_w = 2 \sqrt{A_w / \pi} \). If \( <CT_{ROI} > \) is the mean CT number in a ROI

| Table 1 | Summary of patient data for chest examinations. Weight, height, and BMI are indicated as mean values, with the range in brackets. |
|----------|----------------------------------|
|          | # of Patients | Weight (kg) | Height (cm) | BMI (kg/m²) |
| CT11     | 31 (24 M; 7 F) | 69 (45–105) | 167 (150–183) | 25 (16–38) |
| CT14     | 49 (27 M; 22 F) | 69 (43–117) | 164 (147–183) | 25 (17–53) |
| Total    | 80            | 64 (43–117) | 159 (147–175) | 26 (17–53) |

| Table 2 | Summary of patient data for CAP examinations. Weight, height, and BMI are indicated as mean values, with the range in brackets. |
|----------|----------------------------------|
|          | No. of Patients | Weight (kg) | Height (cm) | BMI (kg/m²) |
| CT11     | 41 (22 M; 19 F) | 65 (44–90) | 163 (145–180) | 25 (17–37) |
| CT14     | 78 (38 M; 40 F) | 70 (43–104) | 164 (144–185) | 26 (17–34) |
| Total    | 119           | 65 (43–90) | 159 (144–175) | 26 (17–37) |
(region of interest) of area, \( A_{\text{ROI}} \), containing the object, then \( A_w \) can be determined from a CT image using eq. (1)\(^{16} \):

\[
A_w = \frac{1}{1000} < CT >_{\text{ROI}}A_{\text{ROI}} + A_{\text{ROI}}
\]

The air surrounding the object should have negligible impact on the result.\(^{16} \) To account for the attenuation of the CT table, \( A_w \) (table) was determined by manually contouring the table (\( \text{ROI}_{\text{table}} \)) in one CT image for each scanner, and then substituting \( A_{\text{ROI}} \), and \( < CT >_{\text{ROI}} \) in eq. (1).

For each examination included in the study, a complete sequence of CT images (image series) was downloaded from the PACS and analyzed using ImageJ software (National Institute of Health, Bethesda, MA, USA), with a macro written by the authors. For each CT image, this macro extracted the values of table position (\( z \)) and tube current (\( I \)) from the DICOM header, then drew a region of interest (ROI) encompassing the entire FOV, and determined its area (\( A_{\text{ROI}} \)) and mean CT number (\( < CT >_{\text{ROI}} \)). The results were transferred to a spreadsheet, and \( D_w(z) \) was calculated using eq. (2)\(^{16} \):

\[
D_w(z) = 2\sqrt{\frac{A_w(z) - A_w(\text{table})}{\pi}}
\]

The automated method to obtain \( D_w \) was tested using two cylindrical acrylic phantoms (32 cm and 24 cm in diameter), water, and imaged with the clinical protocol for chest. The \( D_w \) results obtained were in good agreement (better than 0.2 cm) with expected values.

The values of \( D_w \) obtained with the automated method were compared with \( D_w \) obtained from manual patient contouring in two images (one in the thorax and one in the abdomen) for a total of ten examinations (five in CT11 and five in CT14).

The values of \( D_w(z) \) obtained from patient images were used to calculate two different quantities: the mid-scan or central \( D_w(\text{ave}) \), which is the value of \( D_w(z) \) in the middle of the scanned region; and the \( D_w(\text{ave}) \), calculated as the mean of all \( D_w(z) \) values in the imaged region. \( D_w(\text{ave}) \) and \( D_w(\text{ave}) \) were determined for all examinations, and SSDE was calculated as \( \text{CTDI}_{\text{vol}} \times f(D_w(\text{ave})) \), where \( f(D_w(\text{ave})) \) is the conversion factor related to a patient diameter of \( D_w(\text{ave}) \) obtained from AAPM tables.\(^{15,16} \)

| Table 3 | Acquisition parameters used in both scanners. |
|----------|---------------------------------------------|
|           | kV  | Collimation (mm) | Pitch | Time (s)/rot | Image quality index (NI/SD) | \( I_{\text{min}} \) (mA) | \( I_{\text{max}} \) (mA) |
| CT11      | 120 | 20 (16 × 1.25)   | 1.375 | 0.8          | NI = 18                     | 100                          | 440                          |
| CT14      | 120 | 32 (16 × 2)      | 0.938 | 0.5          | SD = 12.5                   | 80/100\(^a\)                | 500                          |

\(^a\)While data were being collected, the \( I_{\text{min}} \) value was changed for the CT14 scanner. This had little influence on patient doses. Therefore, only one dataset was considered. Image quality remained satisfactory to the radiologists in the department.

3 | RESULTS

3.A | Dosimetric plots as a function of different metrics

Values of \( \text{CTDI}_{\text{vol}} \) are plotted as a function of patient weight (\( a \)), \( D_w(c) \), and \( D_w(\text{ave}) \) in Figs. 1 and 2, for chest and CAP examinations performed in CT11 and CT14, respectively.

The approximate \( \text{CTDI}_{\text{vol}} \) values for different weight ranges listed in AAPM protocols are shown for comparison.\(^{12,13} \) The dispersion of \( \text{CTDI}_{\text{vol}} \) data in Figs. 1(a) and 2(a) reflects the different heights of patients with similar weight, as well as different mass distributions in the body. When \( \text{CTDI}_{\text{vol}} \) is plotted as a function of \( D_w(\text{ave}) \), as shown in Figs. 1(b) and 2(b), there is a separation of data for chest and CAP examinations, related to the different anatomical location of the mid-scan slice. \( D_w(\text{ave}) \) is obtained in the middle of the lungs (low attenuation region) in chest examinations and closer to the liver in CAP examinations.

Plotting \( \text{CTDI}_{\text{vol}} \) as a function of \( D_w(\text{ave}) \) reduces data dispersion to a minimum, as shown in Figs. 1(c) and 2(c). Details become clearer, like the flattening of the curves for very small and very large patient sizes, which is probably related to \( I_{\text{min}} \) and \( I_{\text{max}} \). As expected, the dependence of \( \text{CTDI}_{\text{vol}} \) on \( D_w(\text{ave}) \) is the same for CAP and chest examinations—these examinations are performed with the same ATCM settings, in both scanners.

SSDE values are plotted in Fig. 3(a) as a function of \( D_w(\text{ave}) \) (CAP examinations) and as a function of \( D_w(\text{ave}) \) in Fig. 3(b) (CAP) and 3(c) (chest). The use of \( D_w(\text{ave}) \) reduces data dispersion in plots of SSDE, as it did for \( \text{CTDI}_{\text{vol}} \). To allow a more quantitative comparison, \( R \)-squared values are presented in Table 4 for linear fits to \( \text{CTDI}_{\text{vol}} \) and SSDE data for CAP examinations.

Before the widespread use of ATCM systems, Menke tested different surrogates for mean patient attenuation and concluded that
the correlation between patient attenuation and body mass index (BMI) was no better than with patient weight. A similar result was obtained in this study, as shown in Table 4.

3.B | Mathematically simulated scenarios

In Fig. 4, CTDI\textsubscript{vol} and SSDE are plotted as a function of \(D_{w_ave}\) for the original dosimetric data in CT14 and for the mathematically simulated CAP examinations with \(I_{min}\) values of 140 mA and 180 mA. When \(I_{min}\) is set at 180 mA, CTDI\textsubscript{vol} remains approximately constant for small patients (\(D_{w_ave} < 25\) cm) and then increases gradually until, for large patients (\(D_{w_ave} > 30\) cm), it reaches the values obtained at lower \(I_{min}\) settings. As a result, a high value of \(I_{min}\) reduces the range of variation of CTDI\textsubscript{vol} with patient size. A similar effect is observed for SSDE (Fig. 4).

The mean values and standard deviation (SD) of CTDI\textsubscript{vol} and SSDE are presented in Table 5, for the real examinations and for the simulated scenarios.

3.C | Interdependence of different metrics

The patient sample considered in this work is representative of a particular population of oncological patients (Tables 1 and 2). Mean male and female heights agree well with known statistics for the Portuguese population in this age-group. In Fig. 5, \(D_{w_ave}\) is plotted as a function of patient weight for chest and CAP examinations. The correlations found by Menke for chest and abdominal examinations are shown for comparison. For chest CT, the data from this study fall mostly within the 95% prediction limits previously obtained for this unrelated population (Fig. 5), despite the difference in the obtained regression equation. It is unclear whether this difference results from intrinsic population metrics, or there is some additional bias in this study due to oncological risk factors and/or effect of oncological treatments.
For CAP examinations, the variation of $D_{w,\text{ave}}$ with patient weight is similar to that obtained by Menke for abdominal examinations (Fig. 5).

There is good correlation between $D_{w,c}$ and $D_{w,\text{ave}}$, for both examination types studied, as shown in Fig. 6. The two metrics are very similar, with maximum percentage differences below 15% (in % of $D_{w,\text{ave}}$) as shown in Fig. 7 and summarized quantitatively in Table 6. There appears to be some separation of male and female patients, probably related to differences in body habitus. The mean $D_{w,c}/C0D_{w,\text{ave}}$ difference for chest CT was $4.6\%$, which is comparable with the $1.4\%$ reported by Anam et al.20

The variation of $D_w$ found in each examination, $D_{w,\text{max}} - D_{w,\text{min}}$, is similar for male and female patients, as shown in Fig. 7 and Table 6. The values agree well with those reported by Leng et al. for a different population (weight 37–183 kg, mean 85 kg, BMI 15–57 kg/m², mean 29 kg/m²), where $D_w$ variation was $5.2 \pm 1.4$ cm for chest and $6.5 \pm 1.3$ cm for CAP examinations, and maximum $D_{w,\text{max}} - D_{w,\text{min}}$ was 10.5 cm (32% of $D_{w,\text{ave}}$).21 In this study, $D_w$ variation was $5.8 \pm 1.6$ cm for chest and $6.5 \pm 1.4$ cm for CAP examinations (Table 6).
also allowed retrospective studies and comparisons, for all CT examinations where full FOV images were available.

4 | ANALYSIS AND DISCUSSION

4.A | Dosimetric plots as a function of different metrics

Despite the very similar values of $D_{w,c}$ and $D_{w,ave}$, total examination doses (CTDIs or SSDE) clearly have a stronger correlation with $D_{w,ave}$ than with $D_{w,c}$, as reflected by the lower dispersion of dosimetric data (Figs. 1, 2, and 3, Table 4). The values of $D_{w,c}$ probably reflect both variations in scan length (which alter the anatomical location of the mid-scan slice) and localized anatomy characteristics like abdominal obesity, or large breasts in some female patients. This makes $D_{w,ave}$ the most advantageous metric for the purpose of protocol optimization and automatic detection of outliers or accidental changes, despite the additional computational effort involved in its calculation.

Another advantage of $D_{w,ave}$ is the similar dependence of CTDIvol and SSDE for both examination types (Figs. 1 and 2). Identifying examination type can be challenging for automatic systems, because examination and protocol nomenclature are rarely standardized. Moreover, some examinations have more than one sequence (e.g., before and after contrast injection), making it difficult to compare total DLP. Our data suggest that comparing CTDIvol vs. $D_{w,ave}$ for a group of different examinations (performed with similar protocols) may be a feasible alternative in some situations.

4.B | Mathematically simulated scenarios

The mathematically simulated scenario with $I_{min} = 180$ mA illustrates how patient doses may increase by nearly 50% for adults under $y = 0.1978x + 14.151$

$R^2 = 0.7715$

$15 20 25 30 35 40$

$20 30 40 50 60 70 80 90 100 110 120 130 140$

$D_{w,ave} (cm)$

$W (kg)$

CAP

M

F

All

$y = 0.9383x + 0.6323$

$R^2 = 0.917$

$y = 0.9715x - 0.5063$

$R^2 = 0.9146$

$y = 1.0706x - 0.4339$

$R^2 = 0.8877$

$y = 1.071x + 0.1906$

$R^2 = 0.9033$

$y = 1.0517x - 0.5561$

$R^2 = 0.9435$

$M$

$y = 0.9394x + 1.009$

$R^2 = 0.8225$

$F$

$y = 0.1834x + 14.105$

$R^2 = 0.8225$

$15 20 25 30 35 40$

$D_{w,c} (cm)$

$D_{w,ave} (cm)$

F

M

All

M

F

All

$y = 0.9394x + 1.009$

$R^2 = 0.9453$

$y = 0.9383x + 0.6323$

$R^2 = 0.917$

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$R^2 = 0.9146$
70 kg, while mean examination doses increase by only 12–13%, and maximum values are not exceeded (Fig. 4 and Table 5). This highlights the importance of establishing diagnostic reference levels as a function of patient size, to allow quick detection of nonoptimized protocols by dose monitoring software. Comparison with established references could also be made before irradiation, if \( D_w \) were determined from scout images as suggested by TG220.16

It is not the purpose of this work to determine ideal dose levels for small and large adults, or discuss whether CTDI\(_{vol}\) should increase linearly with patient size. For adult patients, optimum variation of examination dose with size remains a matter of debate. Noise constant systems such as GE and Toshiba result in a linear increase in CTDI\(_{vol}\) with patient weight, but some ATCM systems intentionally decrease dose less for thinner adults.26 A comparison of CT scanners from three manufacturers showed that the Philips system had the least variation of DLP with patient weight, when compared with GE and Siemens.27 Some authors using GE scanners divide adult patients into weight categories.7

More data are necessary, especially as automatic selection of tube voltage may soon be a widespread option as well.28 The example presented here merely highlights the importance of choosing a patient size metric which reduces data dispersion to a minimum, to improve detection of normal trends and outliers.

### 4.C Interdependence of different metrics

The comparisons shown in Fig. 5 are an encouraging result, suggesting that the study of a sufficiently large number of different populations and anatomical regions might provide a conversion between patient weight and \( D_w \), for each examination type. This would allow comparison of newer large-scale data based on \( D_w \) metrics with the existing studies and standards based on patient weight.

As reported by other authors, the impact of \( D_{w, ave} - D_{w, c} \) differences on SSDE values is small,21 because the two metrics have quite similar values. This is reflected in the small data dispersion seen in Fig. 3(b) and 3(c). However, the lower dispersion of dosimetric data when \( D_{w, ave} \) is used as a metric for patient size suggests that the effect of small \( D_{w, ave} - D_{w, c} \) differences is probably amplified by the large variation in CTDI\(_{vol}\) (from ~8 mGy to ~16 mGy, a nearly 100% increase) which occurs for a relatively small increase in
**CONCLUSIONS**

This study highlights the importance of automatic calculation of $D_{w,\text{ave}}$, not just for organ dose estimation as already recommended,\(^{16,18}\) but also to make $D_{w,\text{ave}}$ available to end users as a patient size metric for binning similar-sized patients in RDIM systems.

Despite the small percentage difference between $D_{w,c}$ and $D_{w,\text{ave}} (-4 \pm 4\%$ for chest and $+5 \pm 4\%$ for CAP examinations in this study), both CTD\(_{\text{vol}}\) and SSDE present a stronger correlation with $D_{w,\text{ave}}$ than they do with $D_{w,c}$. Our data suggest $D_{w,c}$ values reflect localized anatomy characteristics. The lower dispersion of dosimetric data obtained with $D_{w,\text{ave}}$ makes it easier to identify trends and outliers. This is useful for ATCM optimization and detection of accidental changes. Use of $D_{w,\text{ave}}$ also reduces dependence on examination type, which may be difficult to identify accurately in large-scale databases. Therefore, when implementing automatic determination of $D_{w,c}$ for SSDE calculations, automatic calculation of $D_{w,\text{ave}}$ should definitely be considered as well, despite the additional computational effort involved.

Use of $D_w$ metrics is not yet widely implemented in CT scanners and RDIM systems, but it is important to acquire baseline data for $D_{w,\text{ave}}$ metrics and to establish comparisons with existing standards based on patient weight. Our experience shows that small-scale studies using $D_{w,\text{ave}}$ metrics are feasible in nonautomated scenarios and may be used initially to acquire baseline data from new and retrospective studies.

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**CONFLICT OF INTEREST**

The authors declare that they have no conflict of interests.

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