Dose-volume product (DVP) as descriptor for estimating total energy imparted to patient undergoing CT examination

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Abstract: The purpose of this study is to expand a descriptor for estimating the total energy imparted to a patient undergoing a CT examination and to investigate its relationship to the currently used descriptor. Estimating the total energy imparted to a patient has previously been characterized by dose-length product (DLP). We propose a descriptor which we call the dose-volume product (DVP), defined as the product of the size-specific dose estimate (SSDE) and the volume irradiated in the patient (V). We also present algorithm to automate the calculation of DVP. There are several steps in calculating the DVP: the first is to contour the patient automatically, the second is to calculate the area of patient in every single slice, the third is to calculate the volume of the radiated part of the patient, the fourth is to calculate the water-equivalent diameter (D_W) automatically, the fifth is to calculate the SSDE, and the last is to calculate the DVP. To investigate the effectiveness of the algorithm, we used it on images of phantoms and patients. The results of this study show that the automated calculations of DVP for both body and head phantoms were in good agreement with theoretical calculations. The differences between them were within 2%. DVP and DLP had a linear relationship with R^2 = 0.971 (slope 1099 cm^2, 95% confidence interval (CI), 1047 to 1157 cm^2) and R^2 = 0.831 (slope 248.6 cm^2: CI, 237.6 to 259.7 cm^2), for thorax and head patients respectively.

***Keywords**: dose-length product (DLP), dose-volume product (DVP), size-specific dose estimate (SSDE), volume CT dose index ($CTDI_{vol}$), water-equivalent diameter (D_W)*

1. Introduction

The patient dose during a CT scan is greater than in other radiological modalities.[1-4] There are at least three reasons for this. Firstly, to get an image of a single slice, the exposure is continuous around the patient, rather than in plain radiography where the exposure is taken from one or two source locations.[5,6] Secondly, the radiation measured along the longitudinal axis appears as a bell-shaped curve and the dose distribution is almost always wider than the beam width. This means that the radiation received by a particular slice of tissue is increased by the radiation received from adjacent slices.[7-9] And thirdly, to get a CT image with an acceptable noise level requires a high tube current (mA).[10, 11] These three reasons result in a dose for CT that is higher than in other modalities, making it very important to estimate CT dose accurately.

Estimating patient dose in a CT examination requires a unique descriptor and associated methodology which differ from those used in other modalities. The descriptor and the methodology change with the development of CT technology. The first descriptor to estimate the dose of a CT scan was the multiple scan average dose (MSAD).[12,13] It was calculated as the average of the multiple scan dose profile (MSDP) along the longitudinal axis for a series of scans.[14,15] The MSDP was measured using thermoluminescent dosimeters (TLD's). These required careful handling and a long acquisition time.

Shope et al.[12] introduced the CT dose index (CTDI) as a very efficient method to estimate CT dose. CTDI measurements needed only a single axial scanning, using a pencil ionization chamber with a length of 100 mm capable of measuring the total dose along a 100 mm longitudinal axis. It used a standard PMMA phantom with a diameter of 32 cm (body phantom) and 16 cm (head phantom).[16] The dose measured with the pencil ionization chamber was then divided by the width of the radiation beam.[12, 16] The CTDI is only the same as the MSAD for a considerable number of scans.[12, 17] Many derivative descriptors based on CTDI, namely $CTDI_{FDA}$, $CTDI_{100}$, $CTDI_W$, $CTDI_{vol}$, were introduced later.

For estimating the total energy imparted to a patient, $CTDI_{vol}$ was multiplied by the scan length, to give the dose-length product (DLP).[18] The DLP reflects the total dose attributable to the complete scan acquisition,[16] and it is proportional to the length of the scanning. For example, an abdomen-pelvis CT examination with the same $CTDI_{vol}$ as an abdomen CT examination would have a greater DLP due to the greater scan length. Changes in scan parameters, e.g. mAs, voltage and pitch, affect the $CTDI_{vol}$ and therefore also the DLP, while a change in scan length affects only the DLP. The DLP is a useful quantity to compare dose levels and it has become accepted through the establishment of diagnostic reference levels (DRL).[19]

The DLP is also very useful in calculating an estimate of the effective dose (ED). The effective dose is considered the most appropriate descriptor to compare patient dose across different imaging modalities.[20,21] For example, the ED value for patients undergoing CT examinations is typically about 5 mSv, while for patients undergoing plain radiography it was typically one hundredth of that, namely around 0.05 mSv.[22] The effective dose is a common method for estimating the risk associated with radiation exposure.[23] The effective dose is calculated by multiplying DLP by the *k* factor, which is widely available and has been extended to cover both sexes, children and newborns.[20] Recently the *k* factor has also been revisited and adapted to modern CT scanners and wide detectors, a range of voltage values, and all ages available in the Oak Ridge National Laboratory (ORNL) family phantom.[24] But effective dose remains a population-based average for patients of standard size.[19,23]

Although $CTDI_{vol}$ and DLP are adopted as descriptors by the practitioners,[16, 25] a number of limitations remain in place. $CTDI_{vol}$ is only calculated for standard phantoms with a diameter of 32 cm to represent the patient's body and 16 cm to represent the patient's head, and the standard phantom was homogeneous.[26] In 2011, the American Association of Physicists in Medicine (AAPM) issued a report that the estimated dose in CT should take into account the size of the patient (effective diameter), leading to the size-specific dose estimate (SSDE).[27] In 2014, AAPM refined the descriptor for characterizing patients using the water-equivalent diameter (D_W), which takes into account both the size and attenuation of patient.[28]

Interestingly, DLP estimates the total energy imparted to a patient by integrating only along the longitudinal axis.[16] In fact, since the radiated object is three-dimensional an accurate estimate must consider all three axes. This study expands a descriptor (from DLP into DVP), presents the algorithm for calculating it, and compares the expanded descriptor to the current descriptor (DLP).

2. Materials and methods

2.1. The dose-volume product (DVP)

Absorbed dose, a physical non-stochastic quantity, is defined simply as the ratio of the energy imparted (*E*) by any ionizing radiation to the matter of mass (*m*). Therefore, the energy imparted can be written as:

$$E = D \times m \tag{1}$$

or

$$E = D \times \rho \times L \times A \tag{2}$$

D is absorbed dose and ρ is mass density of the matter.

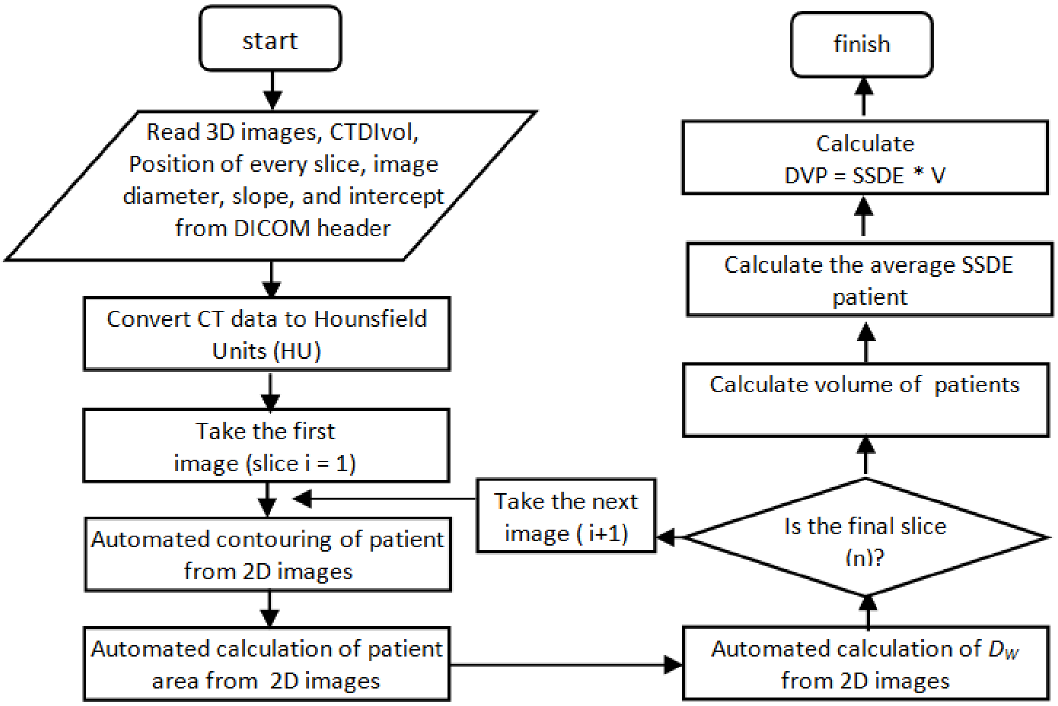

Figure 1. Flow chart of volume and DVP calculations.

The current descriptor for estimating total energy imparted to the patient underwent CT examination is the dose-length product (DLP), calculated as product of $CTDI_{vol}$ and *L*. The unit of DLP is mGy-cm. There are two variables for calculating DLP. The first one is $CTDI_{vol}$ which estimates the dose of the patient. Currently, the descriptor for estimating patient dose is the size-specific dose estimate (SSDE) instead of $CTDI_{vol}$.[27] $CTDI_{vol}$ is considered as an output dose descriptor only and not as a patient dose descriptor.[29] The SSDE is calculated using equation:

$$\mathrm{SSDE} = \mathrm{CTDI}_{\mathrm{vol}} \times k(D_W) \tag{3}$$

where $k(D_W)$ is the conversion factor to convert $CTDI_{vol}$ into SSDE. The $k(D_W)$ is a function of D_W which characterizes the size and composition of patients. The value of $k(D_W)$ is calculated[27] using:

$$k(D_W) = a \times e^{-b \times D_W} \tag{4}$$

where the value of *a* is 3.704369 and 1.874799 for body and head phantom respectively; and the value of *b* is 0.03671937 and 0.03871313 for body and head phantom respectively.[27] D_W in equation (4) is calculated[30] using:

$$D_W = 2\sqrt{\left[\frac{1}{1000}\overline{CT(x,y)} + 1\right]\frac{A}{\pi}} \tag{5}$$

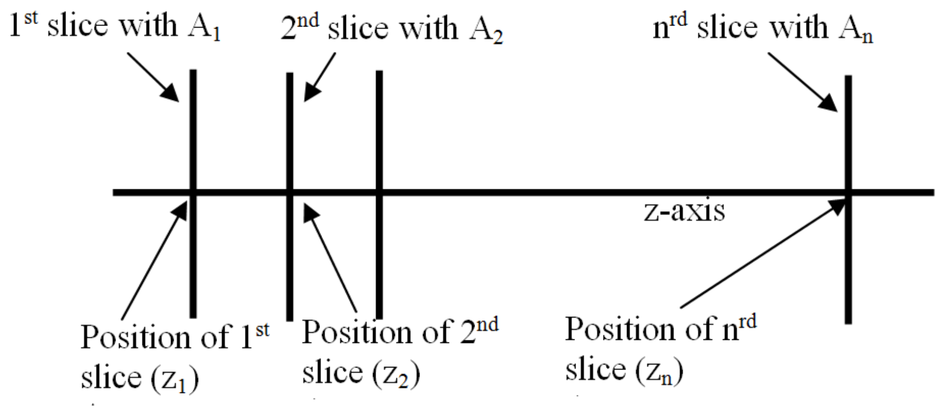

Figure 2. Determination of slice positions.

The second variable in DLP is scan length. Actually, this descriptor neglects the area of the patient in every slice along the longitudinal axis. For a more feasible estimate, we use volume (*V*) instead of length (*L*). Therefore, a better descriptor for estimating the energy imparted to the patient is the product of SSDE and volume (*V*), instead of the product of CTDI_{vol} and length scan (*L*). We refer this descriptor as the dose-volume product (DVP), which is calculated using:

$$\mathrm{DVP} = \mathrm{SSDE} \times V \tag{6}$$

or

$$\mathrm{DVP} = \mathrm{SSDE} \times L \times A \tag{7}$$

The unit of DVP is mGy-cm^3. The main challenge is to automatically and accurately obtain the SSDE and volume of the radiated part of the object (patient or phantom). For automated SSDE calculation, we refer to a previous study.[31] In this study, we present the algorithm for automated calculation of the volume of radiated part of the objects, and the DVP. Using an automated calculation makes this descriptor (DVP) easier to be implemented in the clinical environment.

2.2. The algorithm for DVP calculation

The algorithm for automating the calculation of DVP is shown in Figure 1. The first step is to read the 3D CT images. After that, the values CTDI_{vol}, DLP, position of every slice, slope, and intercept (for converting from CT data into Hounsfield unit) are extracted from the DICOM header. If the CTDI_{vol} and DLP are not available in the DICOM header, these values are recorded manually from the CT console. Next, is to convert CT data into Hounsfield Units (HU) using: HU = CT data × slope + intercept.

The calculation of area and D_W is started from the first slice to the last slice. To calculate both, we must first do automated contouring.[32] From the result of automated contouring, we calculate the area and D_W using equation (5) for every slice. After that, we calculate the volume of the radiated object (patient or phantom) by integrating the area of every slice along the longitudinal axis. Numerical integration is performed using the trapezoidal method:

$$V = \sum_{i=1}^{n-1} \frac{1}{2} d_n (A_i + A_{i+1}) \tag{8}$$

The distance of every slice *d* is calculated using:

$$d_1 = z_2 - z_1 \tag{9}$$

Table 1. Sexes, ages and scan parameters for patients who underwent a thoracic CT examination.

Number Patient	Sex (M/F)	Age (year)	Voltage (kVp)	Tube current (mA)	Exposure time (ms)	Slice thickness (mm)	Total slice
1	M	13	130	72	600	10	34
2	M	58	130	78	600	10	31
3	M	28	130	41	600	10	31
4	F	49	130	69	600	10	29
5	F	64	130	58	600	10	28
6	M	54	130	162	600	10	30
7	M	85	130	113	600	10	34
8	M	42	130	79	600	10	32
9	F	42	130	116	600	10	29

Table 2. Sexes, ages and scan parameters for patients who underwent a head CT examination.

Number Patient	Sex (M/F)	Age (year)	Voltage (kVp)	Tube current (mA)	Exposure time (ms)	Slice thickness (mm)	Total slice
1	M	61	130	167	1500	3	55
2	M	72	130	167	1500	4	24
3	F	52	130	167	1500	4	24
4	M	69	130	167	1500	4	24
5	F	31	130	167	1500	4	24
6	M	58	130	167	1500	4	24
7	F	60	130	167	1500	4	24
8	F	73	130	167	1500	4	24
9	F	46	130	167	1500	4	24
10	F	34	130	167	1500	4	24
11	M	16	130	167	1500	4	27
12	M	36	130	167	1500	4	27
13	F	77	130	167	1500	4	24
14	M	56	130	167	1500	4	24
15	F	34	130	88	1000	4	39
16	M	32	130	167	1500	4	30
17	M	72	130	165	1000	4	20

where z is the position of any particular slice (see Figure 2). The position of every slice was extracted from the DICOM header.

After computing D_W for every slice, the average D_W is calculated. After that we calculate the conversion factor $k(D_W)$ for a specific D_W. Multiplying $k(D_W)$ and CTDI_{vol} gives the SSDE values. And multiplying SSDE and volume gives DVP directly. All of these steps can be fully automated, although in this study we obtained the value of CTDI_{vol} manually from the monitor console because it was not available in the DICOM header.

2.3. Patient and phantom images

In this study, the DVP values were calculated for patients and standard phantoms. Nine patients underwent a thoracic CT and seventeen underwent a head CT were examined. Sex, age and scan parameters are shown in Tables 1 and 2. The DVP values are also calculated for the standard phantoms. The phantoms were scanned using a Siemens Sensation CT scanner. The scan parameters are shown in Table 2. The phantoms were made from PMMA material with ρ = 1.19 g/cm^3, approximately 120 HU, 16 cm in diameter for head phantom and 32 cm in diameter for body phantom. Besides calculating the parameters automatically for the phantoms we also calculated them manually. The volume of the phantom is given by:

Table 3. Scan parameters for head and body phantoms.

Scan parameters	Body Phantom	Head Phantom
Scan type	Helical	Helical
Voltage	120 kVp	120 kVp
Tube current	400 mA	370 mA
Exposure time	500 ms	1000 ms
Slice thickness	3 mm	3 mm
Pitch	1	1
Total of slice	50	50

Table 4. Results of automated and theoretical calculation of $CTDI_{vol}$, L, DLP, D_W, SSDE, V and DVP values for head and body phantoms.

Descriptors	Head	Body
$CTDI_{vol}$ (mGy)	58.02	13.42
L (cm)	14.7	14.7
DLP (mGy-cm)	852.9	197.3
D_W		
Theoretical (cm)	16.94	33.87
Automated (cm)	16.79	33.48
Difference (%)	-0.89	-1.15
SSDE		
Theoretical (mGy)	56.42	14.95
Automated (mGy)	56.77	14.50
Difference (%)	0.62	-3.01
Volume		
Theoretical (10^3*cm^3)	2.89	11.58
Automated (10^3*cm^3)	2.91	11.70
Difference (%)	0.69	1.04
DVP		
Theoretical (10^3* mGy-cm^3)	163.05	173.12
Automated (10^3* mGy-cm^3)	165.20	169.65
Difference (%)	1.32	-2.00

$$V = \pi r^2 d \qquad (10)$$

where r is radius of phantom and d is length of phantom.

2.4. Analysis

The linear regression model was used to estimate the relationship between DVP and DLP, both for head and thoracic patients, and to estimate the slope of the fitted line, along with its standard error and 95% confidence interval (CI). The squared coefficient of determination (R^2) was computed to find the magnitude of association.

3. Results

3.1 The DVP of phantoms

The $CTDI_{vol}$, L, DLP, D_W, SSDE, V and DVP values for the phantoms are shown in Table 4. The $CTDI_{vol}$ value for the head phantom is about 4.5 times greater than the value for the body phantom because the mAs value for the head phantom (370 mAs) is greater than for the body phantom (200 mAs), and the diameter of the head phantom (16 cm) is also less than that of the body phantom (32 cm). For a given CT technique, the patient dose decreases as patient size increases due to the increased attenuation of the incident x-ray beam.[29, 31-34]

The radiated volumes were calculated automatically using equation (8), while the theoretical volume of the phantom was calculated using equation (10). The differences between the automated and theoretical calculations of DVP are within 2%, indicating the accuracy of our proposed algorithm.

From Table 4, although the value of DLP for the head phantom is 4.5 times greater than for the body phantom, the values of DVP for both phantoms are interestingly very similar. Since the volume of the body phantom is about four times that of the head phantom, multiplying SSDE and volume gives similar values for DVP.

3.2 The DVP of patients

The values of $CTDI_{vol}$, L , DLP, D_W, SSDE, V, and DVP for patients who underwent thorax examinations are shown in Table 5 and for patients who underwent head examinations they are shown in Table 6. The values were calculated in the same way as those in Tables 4, except that the values of V were only calculated automatically.

Table 5 shows that the average of the $CTDI_{vol}$ is 5.46 mGy and the average SSDE is 8.70 mGy for the thoracic region. Thus SSDE in the thorax is greater by about 60% than $CTDI_{vol}$. While for the head region, the average of the $CTDI_{vol}$ is 59.54 mGy and the average SSDE is 57.54 mGy, i.e., the SSDE in the head is lower by about 3.4% than $CTDI_{vol}$.

Table 5. The values of $CTDI_{vol}$, L, DLP, D_W, SSDE, V and DVP for patient thorax.

Patient number	$CTDI_{vol}$ (mGy)	L (cm)	DLP (mGy-cm)	D_W (cm)	SSDE (mGy)	V (10^3*cm^3)	DVP (10^3*mGy-cm^3)
1	4.28	33.00	141.24	22.23	7.00	14.14	98.99
2	5.64	29.00	163.56	24.25	8.56	14.68	125.55
3	4.45	30.00	133.50	21.54	7.46	11.24	83.87
4	5.30	28.00	148.40	19.96	9.45	9.54	90.13
5	3.68	27.00	99.36	20.15	6.51	9.30	60.60
6	8.72	29.00	252.88	27.12	11.89	19.03	226.31
7	5.04	33.00	166.32	22.31	8.22	13.85	113.84
8	4.79	31.00	148.49	21.57	8.02	12.40	99.49
9	7.27	28.00	203.56	23.91	11.16	14.13	157.74

Table 6. The values of $CTDI_{vol}$, L, DLP, D_W, SSDE, V and DVP for patient head.

Patient number	$CTDI_{vol}$ (mGy)	L (cm)	DLP (mGy-cm)	D_W (cm)	SSDE (mGy)	V (10^3x cm^3)	DVP (10^3x mGy-cm^3)
1	68.64	15.90	1091.38	17.16	66.13	3.30	218.11
2	59.12	11.75	694.66	18.87	53.46	2.47	132.31
3	59.12	11.75	694.66	16.46	58.62	2.04	119.71
4	59.12	11.98	708.26	19.02	53.17	2.79	148.54
5	59.12	11.75	694.66	17.54	56.02	2.27	127.01
6	59.12	11.77	695.84	17.68	55.69	2.25	125.52
7	59.12	11.75	694.66	17.22	56.81	2.18	123.77
8	59.12	11.74	694.66	15.30	61.37	1.86	114.34
9	59.12	11.75	694.66	16.59	58.32	1.99	115.90
10	56.75	11.95	678.16	15.34	58.82	1.85	108.57
11	59.12	13.56	801.67	16.01	59.69	2.52	150.59
12	59.12	13.57	802.26	17.13	57.03	2.56	145.81
13	56.75	12.17	690.65	15.53	58.38	2.03	118.29
14	59.12	11.76	695.25	17.46	56.22	2.36	132.69
15	60.28	15.20	916.26	16.69	59.22	2.77	164.17
16	59.12	14.75	872.02	21.38	48.40	4.47	216.56
17	60.28	7.60	458.13	16.01	60.86	1.20	72.99

Figure 3 shows the comparison of SSDE and $CTDI_{vol}$ as a function of water-equivalent diameter (D_W). It can be observed that ratio of SSDE to $CTDI_{vol}$ decrease exponentially as D_W increases. SSDE values are higher than $CTDI_{vol}$ in the thoracic region. In the head region, the ratio of SSDE to $CTDI_{vol}$ is approximately 1.0 in the range of D_W between 15 to 17 cm. The patients have D_W values more than 17 cm, resulting in the average SSDE being lower than $CTDI_{vol}$.

The relationship between DLP and DVP in the thoracic region is linear with a slope of 1099 cm^2 (95% confidence interval (CI): 1047 to 1157 cm^2), as seen in Figure 4 (left). The relation between DLP and DVP in the head region is linear with a slope of 248.6 cm^2 (95% confidence interval (CI): 237.6 to 259.7 cm^2), as shown in Figure 4 (right). The coefficient of determination (R^2) of 0.971 in the thoracic region and 0,831 in the head region show that DLP and DVP are strongly associated.

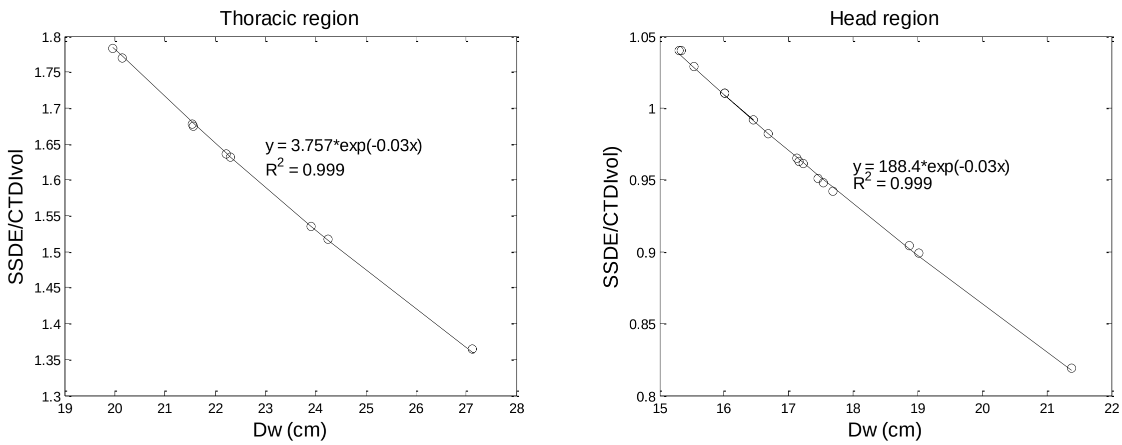

Figure 3. Comparison of SSDE and CTDIvol shown as a function of water equivalent diameter (DW) in the thoracic region (left) and head region (right).

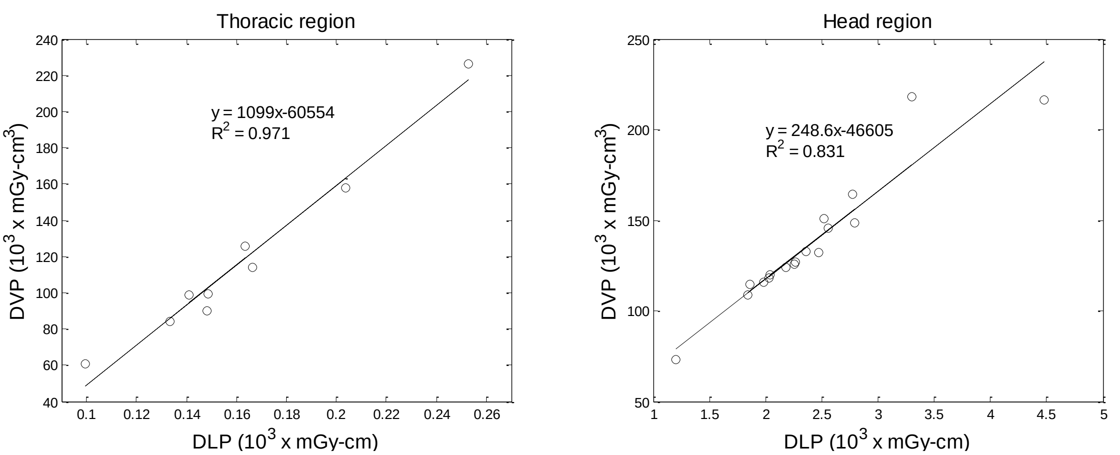

Figure 4. Relationship between DLP and DVP in the thoracic region (left) and head region (right).

4. Discussions

The standard descriptors accepted in the medical practitioners for three decades in CT scan were $CTDI_{vol}$ and DLP. Starting in the mid 2000's, $CTDI_{vol}$ was questioned as a patient dose estimator because it did not take into account the size and attenuation of the patient.[35] There were numerous reports that the radiation dose not only depended on the output of the CT scanner, but also on the size and attenuation of the patient.[32-34,36] To accommodate these concerns AAPM introduced a new descriptor in 2011, called the size-specific dose estimate (SSDE) which took into account the size of patients, characterized by the effective diameter (D_{eff})[27]. In 2014, it updated the descriptor to take into account both the patient size and attenuation, using the water-equivalent diameter (D_W).[28]

Although the descriptor for estimating patient dose is changed, the descriptor for estimating the total energy imparted to the patient is still unchanged, namely the dose-length product (DLP). The DLP is calculated by multiplying $CTDI_{vol}$ and the length of the scan. In this study we expand a descriptor for estimating the total energy imparted which we refer as the dose-volume product (DVP), calculated by multiplying SSDE and volume (V). DVP is an extension of DLP.

We hypothesize that DVP is more feasible than DLP because it is calculated using a more realistic approach, namely using the volume of the radiated part of the patient and the size-specific dose estimates (SSDE) which take into account the output of the CT scanner and the characteristics of the patient in just one direction. Figure 3 shows that the relationship between SSDE and $CTDI_{vol}$ is depend on D_W. In the thoracic region, for this range of D_W, SSDE is always greater than $CTDI_{vol}$, indicating that it is not sufficient to estimate the patient dose using $CTDI_{vol}$. Therefore, it is more feasible to estimate the total energy imparted to the patient by multiplying SSDE and irradiated volume.

We also evaluated the relationship between DVP and DLP for thoracic and head CT examinations. There was a strong linear correlation between them (R^2 were 0.831 and 0.971 for both head and thorax, respectively). Therefore, DVP could be estimated directly using DLP.

The automated calculation of volume is more challenging. In this study we proposed an algorithm for calculating the volume of the radiated object. The algorithm is in good agreement with the manual calculation of the phantom estimate. The difference is within 2%. The accuracy of this algorithm for calculating the volume of the radiated part of the patient still needs to be fully validated.

There were limitations in our study. Firstly, our sample size was small, comprising only nine patients who underwent thoracic CT examination and seventeen patients who underwent head CT examination. Secondly, the volumes of radiated objects were calculated by integration of the area of every slice with d values extracted from the DICOM header. These results are likely underestimates, because the slice position was obtained from the center of the slice. In this estimate, parts of the objects are neglected, namely half of the slice thickness in the left side of the first slice and half of the slice thickness in the right side of the last slice. For a better estimate the missing part of objects must be included. The missing part can be calculated by adding half area of the first slice and half area of the last slice. Fourthly, the total energy imparted to the patent (DLP) is usually used to estimate the organ dose and effective dose (ED). In this study, we have not calculated the effective dose yet. In our next study we will compare the effective dose calculated using DLP and DVP.

5. Conclusion

We have proposed and tested a descriptor for characterizing the total energy imparted to the patient in terms of the dose-volume product (DVP). This descriptor is more realistic for estimating the total energy imparted to the patient than the current descriptor (DLP). The calculation of DVP is straight forward and fast, so it is very practical to implement in the clinical environment. The potential and advantages of DVP need to be investigated further.

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