

# Understanding CT Dose Display

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## INTRODUCTION

CT is responsible for more net radiation exposure to the US population than any other diagnostic test. As fears over medical radiation exposure grow, so too does the importance of understanding CT dose. For example, the California state legislature recently passed a law requiring that CT dose be included in all radiology CT reports beginning July 1, 2012 [1]. It is thus essential for contemporary radiologists to be comfortable accessing and discussing such information. Fortunately, this information is now readily available and contained within the prescan and postscan display modes of recently manufactured CT scanners.

## WHAT EXACTLY ARE VOLUME COMPUTED TOMOGRAPHIC DOSE INDEX AND DOSE-LENGTH PRODUCT?

The two most fundamental units of radiation dose for CT are the volume computed tomographic dose index (CTDI<sub>vol</sub>), expressed in milligrays, and the dose-length product (DLP), expressed in milligray-centimeters. At the most basic level, CTDI<sub>vol</sub> is a measure of the average dose absorbed by a given phantom at a given scanner output (ie, set of scanning parameters or “protocol”). Dose-length product, on the other hand, is normalized for the length of the scan and is thus an estimate of the total absorbed dose.

Values of CTDI<sub>vol</sub> for each protocol are determined by a radiation physicist whenever a new scanner arrives or an existing scanner undergoes major repairs or during periodic quality control assessment [2,3]. An acrylic phantom is placed at the isocenter of

CT gantry rotation to mimic the scatter properties of human tissues. Different phantoms are used for different applications; for example, the standard body phantom is 32 cm in diameter, whereas a 16-cm phantom is used to mimic either the adult head or the pediatric body, and a 10-cm phantom is used to mimic the pediatric head. Ionization chambers are used to measure radiation exposure in different parts of the phantom, after which a standard formula is applied, ultimately yielding CTDI<sub>vol</sub> in milligrays. Dose-length product is equal to CTDI<sub>vol</sub> multiplied by the length of the area scanned in centimeters and is expressed in milligray-centimeters. Dose-length product is especially useful when attempting to convert exposures from the CT dose index (CTDI) paradigm to effective dose in sieverts, the universal currency of patient radiation exposure that encompasses the risk for harm from radiation effects models.

## LIMITATIONS OF COMPUTED TOMOGRAPHIC DOSE INDEX AND DOSE-LENGTH PRODUCT

Considering the strong emotions involved with radiation exposure and the numerous metrics of radiation dose reported in the scientific literature, it is equally important to emphasize what CTDI<sub>vol</sub> and DLP do not represent.

Most important, neither CTDI nor DLP represents the dose absorbed by any specific patient. That is, neither is based on any measurement performed while the patient is in the scanner. Instead, both measures provide standardized estimates of average doses to cylindrical

phantoms that are useful for comparing average doses between protocols and across imaging centers. In other words, they are highly precise, but not necessarily accurate when used to estimate the dose to an individual patient. For example, the uncertainty inherent in translating phantom dose measurements to effective dose for individual patients can be as high as 50% [4]. When these effective doses are further translated to individual cancer risk estimates, further uncertainty is introduced.

An additional caveat involves the importance of the phantom used to make the measurements. Even if all scan parameters are held constant, using different phantoms of different size yields different CTDI<sub>vol</sub> measurements. Because of soft tissue beam scattering, smaller diameter patients and phantoms will absorb more radiation per unit volume. If a patient is significantly larger or smaller than the phantom used to determine CTDI<sub>vol</sub>, the measurement will be inaccurate. This issue is most important with regard to pediatric patients who are significantly smaller than the phantoms available at a given institution and who have a higher risk for radiation-induced cancer per exposure unit. Simple and easy-to-use methods for addressing this problem have been described elsewhere by Strauss and Goske [5].

Last, it is important to recognize that measurements of CTDI<sub>vol</sub> and DLP are not equivalent to effective dose and cannot be additively combined in all cases. For example, a patient who has received a DLP of  $X$  from a head CT scan as well as a DLP of  $Y$  from an abdominal CT scan does not necessarily have a net DLP of  $X + Y$ . Instead, the DLP of

each scan must be converted into an effective dose separately to account for the radiosensitivity of the organs likely to have been irradiated when scanning each body part. It is only after conversion to effective dose that exposures from separate scans can be additively combined. A notable example is when the same body part is scanned in the same session; in this case, it is reasonable to combine DLP (but not CTDIvol) additively.

### PRESCAN DISPLAY

Since 2002, it has been a requirement of the International Electrotechnical Commission to display the CTDIvol for any planned protocol before initiating the scan [6]. Although the display layout differs among manufactures, expression of CTDIvol is universal because of these rules. Figure 1 shows the prescan display from one CT manufacturer (Toshiba Medical Systems, Nasu, Japan), which shows both the calculated CTDIvol and, importantly, the phantom used to determine CTDIvol.

In practice, this information is most useful in ensuring that there are no equipment malfunctions or faulty protocol settings. It can also be used to provide a prescan exposure estimate by obtaining CTDIvol information and multiplying this by an estimate of the proposed scan length to obtain DLP. If more precise information is desired or if the questions arise immediately before the scan, when the patient is “on the table,” a more accurate estimate of DLP can be used on the basis of the projection or “scout” scan (keeping in mind that the scout scan is associated with its own small radiation dose).

An important caveat here involves protocols using automatic exposure control (AEC). With AEC, the scanner modifies the dose in real time on the basis of patient anatomy or cardiac motion. However, the prescan information displays only an average CTDIvol on the basis of phantom measurements. It is not possible to know prospectively how auto-

matic dose modulation will function during a scan, which introduces additional uncertainty into the prescan CTDIvol estimate.

Finally, with multiple scans performed (eg, noncontrast, arterial phase, venous phase), it is important to keep in mind that each scan may have a separate CTDIvol and potentially a separate scan length, introducing further uncertainty in the prescan dose estimate.

### POSTSCAN DISPLAY

Once the examination has been performed, more information is available, namely, the number and type of different scans performed and the scan length of each. For most manufacturers, this information is tabulated in the manner shown in the scanner output (Siemens Medical Systems, Erlangen, Germany) in Figure 2, with a row for each scan and a column for each summary statistic. Although the number and type of summary statistics vary by manufacturer, once again, the most important metrics are CTDIvol and DLP.

The postscan CTDIvol represents the value based on the scanner settings used for the scan, with some manufacturers including a column to indicate which phantom was used. For protocols using AEC, this represents an average CTDIvol across the scanned field. Specific information on which organs were scanned at higher or lower CTDIvol due to AEC is not currently available.

The postscan DLP represents CTDIvol multiplied by the actual scan length used. As discussed above, one must be careful when attempting to combine information between different lines; only when the same exact anatomy has been scanned multiple times can the DLP be combined additively.



**Fig 1.** Sample prescan dose display at CT control console displaying volume CT dose index (CTDIvol), dose-length product (DLP) for the chosen protocol, and the type of phantom used in determining CTDIvol (Toshiba Medical Systems, Nasu, Japan).

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Ward: NEL-4  
Physician:  
Operator:

Total mAs 4212 Total DLP 968 mGycm

Scan	kV	mAs / ref.	CTDIvol* mGy	DLP mGycm	TI s	cSL mm
Patient Position H-SP						
Topogram	1	120 35 mA	0.14 L	8	5.9	0.6
PreMonitoring	2	120 30	1.79 L	2	0.5	10.0
Contrast						
Monitoring	3	120 30	5.36 L	5	0.5	10.0
Arterial	6	120 170 / 290	11.49 L	333	0.5	0.6
Venous	7	120 184 / 290	12.45 L	620	0.5	0.6

  

Medium	Type	Iodine Conc. mg/ml	Volume ml	Flow ml/s	CM Ratio
Contrast		0	0	0.0	100%
Saline			0	0.0	

\* L = 32 cm, S = 16 cm

**Fig 2.** Sample postscan CT dose report displaying CT scan series and corresponding parameters (Siemens Medical Systems, Erlangen, Germany).

### GUIDELINES FOR EFFECTIVE PATIENT COMMUNICATION

Patients are typically unfamiliar with the units of radiation dose and are instead interested in what harmful effects the scan may have on their health. Because the dose ranges associated with diagnostic imaging are in the stochastic range, it is reasonable to limit the discussion of risk solely to the issue of radiation-induced cancer. With this consideration comes the inevitable questions of how explicit the radiologist should be, how much detail is desirable, and in what form.

It is certainly possible, although somewhat laborious, to convert doses expressed in DLP into effective doses in millisieverts, which can be used to obtain a lifetime excess cancer risk due to the scan. However, cumulative effective dose estimates are inaccurate when used to calculate individual cancer risk and, more important, are not relevant when considering the risks and benefits of future scans [7]. It seems more appropriate to make the default mode of communication something less incendiary but still meaningful to patients, such as comparison with natural back-

ground radiation at sea level or the relative radiation levels used as part of the ACR Appropriateness Criteria® [8]. Offering the information thusly both increases its accuracy and decreases the probability that irrational radiation fears will prevent a medically necessary examination from taking place in the future.

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