THE HISTORY AND EVOLUTION OF CT DOSIMETRY

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Abstract- **A historical description of the development of CT dosimetry and its evolution; including the flaws in the present-day dose-descriptors which have not kept pace with modern CT techniques, and the required modifications for same - which corrections can be applied by the medical physicist.**

I. INTRODUCTION

The following historical vignette lends some perspective to the development of CT dosimetry. This material has been excerpted from my recent Book: **The Physics of CT Dosimetry**, CRC Press.

The early workers referenced here could not have imagined the explosive growth in CT methodology which would occur over the ensuing decades.

II. THE EARLY UNIVERSE

The early measurement of CT dose and mapping of the dose distribution was primarily done using Thermoluminescent Dosimetry (TLD) which was tedious and had relatively low spatial resolution. In the early days of CT when scan times were slow and x-ray tube heat capacities were low, obtaining the dose (or dose distribution) resulting from multiple axial slices was difficult. Ed McCollough and Tom Payne (beginning in 1976) did some early work using TLD.

In 1977, the pencil chamber method was introduced by Jucius and Kambic – the same year the Apple II computer was released, and people were playing the Atari video game, PONG.

Bob Jucius and George Kambic of Ohio Nuclear, Inc. (a US CT manufacturer) provided the first comprehensive look at CT dosimetry, presenting various options including TLD *as well as the introduction of the long pencil ion chamber* which they commissioned Capintec, Inc. to manufacture for them [1]. They derived an equation which showed that the integral of a single slice dose profile could be used to *predict* the average dose about the central scan location ($z = 0$) for multiple slices. This is far from obvious, and their insight was quite impressive. Their derivation involved a (relatively opaque) summation of integrals. They also mapped dose distributions using TLD and surface dose using Kodak RP/M (mammography) film, but concluded that "at this time, TLD is the technique of choice".

Dixon and Ekstrand [2] independently introduced surface dose mapping using a slower radiation therapy verification film (Kodak Xomat /V), digitized using a scanning densitometer for various scanners of the day (resulting in some unexpected dose spikes).

III. THE BIRTH OF CTDI – 1981

Perhaps the best-known paper was that of a US FDA group Shope, Gagne, and Johnson [3] who refined the integral concept of Jucius and Kambic described above. To avoid confusion we will henceforth adopt the following simplified notation used in our Medical Physics Publications and in my Book [4]. Shope et al. defined the "Multiple Slice Average Dose" (MSAD) resulting from a series of N identical axial dose profiles $f(z)$ spaced at equal intervals of $b = \Delta d$ along z as

$$MSAD = D_L(0) = \frac{1}{b}\int_{-L/2}^{L/2} f(z')dz' \qquad (1)$$

Where the MSAD is the average dose over $\pm b/2$ about $z = 0$ (at the center of the scan length L) and where $L = Nb$ (the integration limits and the divisor b are necessarily coupled). For axial scans ("step and shoot") the dose distribution over the scan length is quasi-periodic of period b, hence the average is over one period ($\pm b/2$) about $z = 0$. Note that their nomenclature "multiple scan average dose" (MSAD) is rather misleading, since it is not the average dose over the total scan length, but rather only *about the center of the scan length* $z = 0$. They also stated that L in the above MSAD equation was intended to be long enough for the dose at the center of the scan length to reach its limiting, *equilibrium value*. From this they defined a "dose index" CTDI as

$$\mathrm{CTDI}_\infty = \frac{1}{T}\int_{-\infty}^{\infty} f(z')dz' \qquad (2)$$

where T is "the slice thickness as stated by the manufacturer" and $f(z)$ is the *dose profile* generated by a single axial scan centered at $z = 0$. This is the value of MSAD when L is large enough such that MSAD approaches its limiting (equilibrium) value (which we denote by D_{eq}) – such that profiles beyond $z = \pm L/2$ contribute negligible scatter back to $z = 0$; $z = 0$ being the relevant location for MSAD or CTDI. Note also that $CTDI_\infty$ represents the dose that accrues at the center of the scan length for a table increment $b = T$, which represented "contiguous axial scans". With the advent of multi-detector

CT (MDCT), *T* is replaced by "N x T" (*nT* in our more concise notation used herein). A common misconception is that *T* or *nT* represent a beam width, but physically (in any valid dose formula) they represent a table increment, as illustrated by our derivations of same [4,5].

The derivation of the MSAD equation by Shope and Gagne [3] involved a tedious summation of integrals (following Jucius and Kambic). The derivation for axial scans has been simplified to a few steps [5] using convolution mathematics; this derivation produces the "running mean" dose $D_L(z)$ as an average over $z \pm b/2$ *at all values of z* (and not just $z = 0$ as for the MSAD of Shope et al.). This derivation is shown in Chapter 2 of the Book [4].

IV. ENTER THE REGULATORS (1989)

Codification of physical law rarely turns out well, and once the law has been laid down it is devilishly hard to change (also "too many cooks spoil the broth").

The original definition of CTDI put forth by Shope et al 1981, as well as the original US FDA regulatory proposal [6], used the *infinite* line integral of the single-slice, axial dose profile $f(z)$, viz. $L \rightarrow \infty$ with $b = T$. The meaning and intent of "infinity" were clear and unambiguous to the physicists, symbolically indicating that the integration limits ($-L/2$, $L/2$) must be at least large enough to encompass the complete width of $f(z)$ including its long scatter tails, such that any further increase in L would provide a negligible additional contribution to the accumulated dose at $z = 0$ for a scan length L. This in turn assured that the CTDI, thus defined, would represent the *maximum limiting value* of the accumulated dose at the center of the scan length resulting from multiple, contiguous ($b = T$) scans, namely, *the equilibrium dose* D_{eq}. Had the FDA retained it as originally proposed, it would have been self-correcting and "bullet proof", since many of the ensuing difficulties with CTDI were produced by attempting to define suitable, *finite* integration limits.

But alas, "infinity" did not survive the transformation to the "final FDA rule" (due to public comment; and perhaps because the concept of "infinity" is not in the legal lexicon); and thus the $\pm 7T$ integration limits were adopted - which length the FDA stated [6] "*would produce little difference from the originally- proposed infinite integral for the largest slices then available*" (T = 10 mm), and "*would be representative of typical clinical scan lengths of 10 -15 T.* " (100 – 150 mm). In hindsight, both conclusions were flawed and rapid technological advances led to typical body scan lengths of 250 mm or greater. The FDA did, however, retain the required coupling between the integration limits and the divisor *T*.

A. *The Standard Dosimetry Phantoms*

FDA [6] defined "standard dosimetry phantom" as a right circular cylinder of polymethl-methacrylate (PMMA) of diameters of 32 cm (body) and 16 cm body (head) 14 cm in length which can accommodate a dosimeter both along its axis of rotation and along a line parallel to the axis of rotation 1.0 centimeter from its surface. This truncated length gives a shortfall of $CTDI_{100}$ of 7% on the central axis and 1.3% on the peripheral axes due to missing scatter in a 15 cm long phantom [6].

V. THE QUIESCENT PERIOD

Nevertheless, a long period of quiet acceptance prevailed, during which time the mathematical theory behind the pencil chamber and subscripted CTDI methodology was forgotten (many likely had not even seen the derivation) – and some began to believe that they were making an actual "dose" measurement with the pencil chamber. One does not, and cannot, directly measure a dose with a pencil chamber. Not even in air. Among other things, a pencil chamber reading defies the inverse square law ($1/r^2$). Its reading varies as $1/r$. Many "unwary" diagnostic physicists have fallen into the trap of using the pencil chamber outside of its limited, approved use; supporting the old adage "*if the only tool you have is a hammer, you tend to treat everything as if it were a nail*". The pencil chamber measures *a dose-integral* in units of *mGy.cm*; so even though your electrometer may read *mGy* (or mR) it is likely not programmed for a pencil chamber (and is actually only measuring the charge collected in Coulombs). See [4,7] for pencil chamber calibration methods and units.

VI. ENTER $CTDI_{100}$ - 1995

$CTDI_{100}$ (based on a 100 mm long pencil chamber measurement) was introduced [8] around 1995 as a *more practical* indicator of patient dose, and then widely adopted (based on a European Commission Study Group 1998). The widespread use of the 100 mm chamber seems to have been an *ad hoc* decision, and not supported by the physics. The FDA kept the required coupling between the integral divisor and the integration limits; but variable integration limits were not practical for the pencil chamber methodology. However, a fixed integration length can (and does) lead to anomalies.

Since $CTDI_{100}$ has a different value for the central and peripheral phantom axes, a desire to have a single CTDI number (dose index) to represent "dose" for a national survey in Sweden [8] led to an approximate "weighted average" dose across the central scan plane at $z = 0$ assuming an *ad hoc* linear variation of $CTDI_{100}$ from the central phantom axis to the peripheral axis (p) namely,

$$CTDI_w = \frac{2}{3} CTDI_{100}(p) + \frac{1}{3} CTDI_{100}(c) \qquad (3)$$

The (1/3, 2/3) weighting proves adequate for $CTDI_{vol}$ (based on $CTDI_{100}$); however, the central axis to peripheral axis dose ratio increases as scan length increases beyond

100 mm due to increased scatter thereon. We also note that the actual dose curve $D(r)$ is not linear, but is sigmoidal, with *zero slope* on the central axis ($r = 0$) and again near the phantom surface.

VII. THE ADVENT OF MULTIDETECTOR CT (MDCT) - 1998

The divisor of the CTDI integral now becomes nT (or "N x T") which is the *active detector length* as projected back to scanner isocenter, and represents the total available scan width for reconstruction. The actual primary beam width (*fwhm*) $a > nT$ is required to keep the penumbra beyond the active detectors, called "over-beaming". MDCT allowed reconstruction of smaller slices than nT but with a concomitant increase in noise, e.g., an acquisition using nT = 20 mm, can be reconstructed as four 5 mm slices.

VIII. ENTER $CTDI_{vol}$ (A MISNOMER) BUT AN IMPROVEMENT SINCE IT ELIMINATES nT (N x T)

$CTDI_w$ was later modified by the IEC in 2001 to include the effect of "pitch" (table increment b) on dose as

$$CTDI_{vol} = p^{-1}\, CTDI_w \qquad (4)$$

where $p = b/nT = \Delta d/nT$ applies to both helical and axial scans. The nomenclature $CTDI_{vol}$ is again a misnomer since it does not represent a volume average as its subscript might imply- no average having been taken over the 100 mm scan length; rather it still represents *the planar average dose over the central scan plane* (at $z = 0$) for a 100 mm scan length. Its basis is still $CTDI_{100}$ which is hidden. We also note that *nT cancels out in* $CTDI_{vol}$ such that *only the inverse of the table increment per rotation* b^{-1} *matters* – the divisor nT in $CTDI_{100}$ serves only as a place-keeper.

As the table increment $b \rightarrow 0$, then $CTDI_{vol} \rightarrow \infty$; however, this is nonsensical since the actual dose remains finite. The oft-forgotten required coupling of scan length $L = Nb$ and table increment b in Eq. (1) also requires the integration limits to approach zero, resulting in the dose approaching the eminently-plausible value $Nf(0)$ where N = number of rotations; i.e., the N dose profiles $f(z)$ simply pile up on top of each other at $z = 0$, and $CTDI_{vol}$ (calculated from $CTDI_{100}$) *no longer has any relevance*. This is shown mathematically in [9] as well as Chapter 5, Dixon 2019 [4] for stationary table CT, although it is fairly obvious.

IX. DOSE LENGTH PRODUCT

DLP = L X $CTDI_{vol}$ is a *measure* of the total energy deposited in the phantom. Note that DLP does not depend on the scan length L per se′ since $L = Nb$ and $CTDI_{vol}$ is proportional to b^{-1}; thus b cancels in the product, and DLP really depends only on the number of rotations N or total mAs. Increasing scan length L by increasing pitch alone does not change DLP. Even if the table translation is slowed to a stop ($L \rightarrow 0$) DLP remains the same. DLP is by no means *equal* to the total energy deposited since $CTDI_{vol}$ is based on $CTDI_{100}$ – the total energy deposited is calculated in Chapter 2 in Dixon's book[4]. DLP remains robust for shift-variant techniques, whereas $CTDI_{vol}$ is not.

X. HELICAL SCANNING - SCANNING WITH CONTINUOUS TABLE MOTION - 1990

Willi Kalendar [10] introduces helical scanning ("spiral CT"). Dixon [5] in 2003 derived the dose equations for helical scanning for the *dose* $D_L(z)$ over the entire scan length L, for both the central phantom axis and likewise for the peripheral axis where an angular average over 2π at a fixed value of z is used. This derivation treats the dose rate profile as a traveling wave in the phantom (and is accomplished in a few steps for the central axis on which the dose rate is constant) and is given by the form of a traveling wave $\dot{f}(z - \upsilon t) = \tau^{-1} f(z - \upsilon t)$ where $f(z)$ is the single-rotation (axial) dose profile acquired with the phantom held stationary, υ is the table speed, τ is the gantry rotation period (in sec), and t_0 is the total scan time as illustrated in Fig. 1

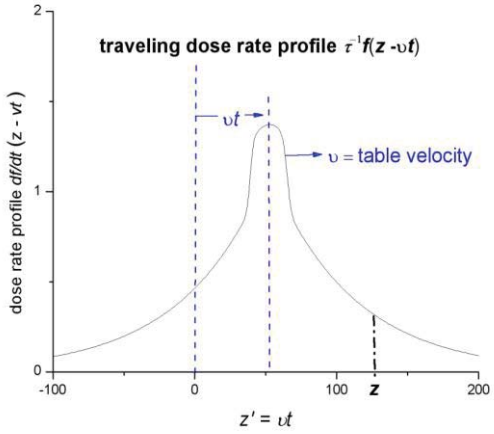

Fig. 1. A traveling *dose rate profile* $\dot{f}(z - \upsilon t) = \tau^{-1} f(z - \upsilon t)$ in the phantom reference frame is created when an axial dose profile $f(z)$ is translated along the phantom central axis z by table translation at velocity υ, where τ is the gantry rotation period (in sec), which has the familiar form of a traveling wave (z' in mm). Note the long scatter-tails on the dose profile in Fig. 2.1 such that the point z will begin accumulating dose long before the primary beam component of width a = 26 mm (nT = 20 mm) has arrived and long after it has passed.

Integrating the dose rate over the total scan time t_0 gives

$$D_L(z) = \tau^{-1} \int_{-t_0/2}^{t_0/2} f(z - \upsilon t)\, dt \qquad (5)$$

$$D_L(z) = \frac{1}{b} \int_{-L/2}^{L/2} f(z - z')\, dz' \qquad (6)$$

$$D_L(z) = \frac{1}{b} f(z) \otimes \Pi(z/L) \qquad (7)$$

the conversion from the temporal to the spatial domain in Eq.(6) having been made using $z' = vt$, scan length $L = vt_0$, and a table advance per rotation $b = v\tau$, resulting in the above convolution in Eq.(7) describing the total dose $D_L(z)$ accumulated at any given z-value during the complete scan, expressed as a convolution with the rect function $\Pi(z/L)$. This reduces to the $CTDI_L$ equation by setting $z = 0$ with a *table increment* $b = nT$, i.e.

$$D_L(0) = \frac{1}{b} \int_{-L/2}^{L/2} f(z')dz' = \frac{nT}{b} CTDI_L \qquad (8)$$

where b/nT is equal to the helical pitch. When $CTDI_L$ is arbitrarily truncated to a scan length of L = 100 mm, it becomes $CTDI_{100}$..

The same equation for $D_L(z)$ was also shown by Dixon [5] to also apply to axial scanning when a longitudinal "running mean" (average over $z \pm b/2$) is used, which also reduces to the *CTDI* paradigm at $z = 0$ as previously discussed. This derivation is likewise shown in Chapter 2 of Dixon 2019 [4], and is easily accomplished using convolution mathematics (as opposed to the tedious summation of integrals previously used by Shope et al.[3] to calculate MSAD and CTDI).

We also note from the derivation, that *the integral format of CTDI devolves from the motion of the phantom,* and that it does not apply to a stationary patient support technique such as use of a wide cone beam without any table motion; and likewise, *a pencil chamber acquisition of the integral* to compute $CTDI_{100}$ has no relevance or utility to such stationary table techniques.

XI. SLIPPING THE SURLY BONDS OF CTDI

The CTDI-paradigm has many limitations which are not widely-appreciated as described in this section. The CTDI-paradigm requires *shift-invariance* for which no scan (or phantom) parameters can vary with *z*, therefore it cannot apply to many modern *shift-variant* CT techniques such as tube current modulation (TCM). It also only applies to phantom-in-motion techniques, and not to stationary patient-support protocols.

A. *An Alternative To The Pencil Chamber – 2003*

Dixon in his 2003 paper [5] also described an alternative measurement method to that of the pencil chamber of fixed length which is much more versatile. Unlike early CT scanners, modern CT scanners can scan over any desired length of phantom in a few seconds, therefore integrating the dose from a small ion chamber fixed in a moving phantom can give the accumulated dose for any scan length or clinical protocol, and thus can emulate a pencil chamber of any arbitrary length (and can even be used to measure $CTDI_{100}$). That is, the small ion chamber can be used in this way to create a "virtual pencil chamber" of any desired length. This method has been validated experimentally in detail in Dixon-Ballard [7] and is also described in Dixon Chapter 3 [4] where a 0.6cc Farmer ion chamber is shown to give the same result as a 100 mm and 150 mm pencil chamber – and for any other scan length *L* as well. It is also immune to the *shift-variant* problems discussed below.

B. *AAPM TG-111 - 2010*

A Task Group of The American Association of Physicists in Medicine published AAPM Report 111 [11] entitled "Comprehensive Methodology for the Evaluation of Radiation Dose in X-ray Computed Tomography" in which the small ion chamber is utilized for measurements rather than the pencil chamber, and which recommends a return to the equilibrium dose D_{eq} as the measurement goal (as originally recommended by Shope et al. 1981 [3] and the FDA[6]). There is no mention in this report of CTDI nor the pencil chamber.

C. *Limitations Of The CTDI-Paradigm And The Pencil Chamber Acquisition.*

The CTDI-paradigm has significant limitations. It only applies to moving patient-support techniques, such as helical scanning or an axial scan series, as discussed above. Every dose profile $f(z)$ in such a scan series must be identical to that integrated by the pencil chamber in order for *the predictive method* of CTDI to be valid; in other words, it requires *shift-invariance* for which no scan parameters can vary with *z*. That is, it requires constant tube current (mA), constant pitch (or table increment *b*), and a constant phantom cross-section along *z*. Therefore, it cannot apply to Tube Current Modulation (TCM) which is commonly-utilized today. Dixon and Boone [12] derive the proper dose equations for such *shift-variant* techniques (TCM and pitch modulation) shown in Chapters 7 and 8 of Dixon 2019 [4] as well as in [13] and [14].

The small ion-chamber method has no such restrictions. It can even be deployed in an anthropomorphic phantom. It is measuring an actual accumulated dose, and not relying on the predictive methodology of CTDI, which uses the integral of a single scan to *foretell* the dose at the center of the scan length which would accrue if *identical* scans were laid down at equal intervals over, a 100 mm scan length as for $CTDI_{100}$ and thence for $CTDI_{vol}$.

XII. THE IEC ATTEMPTS TO CIRCUMVENT THE LIMITATIONS OF CTDI

"If the only tool you have is a hammer, you tend to treat everything as if it were a nail"

$CTDI_{100}$ (thence $CTDI_{vol}$) does in fact have a precise physical meaning: it is equal to the actual accumulated dose in-phantom at the center of a series of contiguous scans ($b = nT$) *covering one specific scan length, L = 100* mm; but it

underestimates the limiting equilibrium dose D_{eq} (as well as the accumulated dose for any scan length above 100 mm) - particularly for typical clinical body scan lengths of 250 – 500 mm which approach the equilibrium dose. It also *over-estimates* the dose for $L < 100$ mm.

The IEC [15] has attempted to "prop-up" $CTDI_{vol}$ and its "hand maiden" the 100 mm long pencil chamber, in a series of patches. *These patches govern the scanner-reported* $CTDI_{vol}$, as discussed below.

A. *For shift-variant techniques* such as TCM, the IEC version uses the average of $mA(z)$ *over the entire scan length* as if it were a constant mA in the CTDI-paradigm; whereas $CTDI_{vol}$ *applies only to a 100 mm scan length* – a clear disconnect. This creates a "$CTDI_{vol}$ of the second kind" and the disconnect negates a possible physical interpretation of "$CTDI_{vol}$ (TCM) as illustrated in Chapter 7 in Dixon's book[4]. IEC also introduces the absurdities which are supposed to represent local doses: $CTDI_{vol}(z)$ and $CTDI_{vol}(t)$; but which (apart from having units of dose) are not doses at all, but merely surrogates for $mA(z)$ as likewise shown in [4]. The *local dose* at z does not track $mA(z)$ [or $mA(t)$] since it also consists of scatter from the entire scan length – to paraphrase Charles Dickens "local dose also depends on "mA past and mA yet to come". See Fig. 1 in which the height of the traveling profile for TCM now varies with time or $z' = vt$. The correct equation for TCM derived in Chapter 7 Dixon 2019 [4] is given by

$$\widetilde{D}_L(z) = \frac{1}{b}\int_{-L/2}^{L/2} i(z')\,\hat{f}(z-z')dz' \qquad (9)$$

$$\widetilde{D}_L(z) = \frac{1}{b}\,\hat{f}(z)\otimes\left[i(z)\Pi(z/L)\right] \qquad (10)$$

in which the tube current at all locations z' along z contributes to the dose depending on the magnitude of the scatter tails of the axial dose profile $f(z)$ per unit current at z', via a convolution with the $i(z)$ = mA(z) profile in brackets; rather than a direct product as the IEC definition would imply (the latter being tantamount to removing $i(z')$ from the integral and replacing it with its average value over L – not to mention truncating the integral to 100 mm). In point of fact, for a scan length of 100 mm, fully 44% of the energy deposited about the central phantom axis is deposited *outside the scan length* where mA(z) = 0 (Table 7.1 Dixon 2019[4]); and where $CTDI_{vol}(z)$ likewise drops to zero although the actual dose does not.

B. *Stationary phantom/table*

For the stationary phantom/table to which the CTDI-paradigm does not apply, the IEC solution is $CTDI_{vol} = N$ x $CTDI_w$ where N is the number of rotations. Its failure by up to 300% for narrow beam perfusion studies and for wide cone beams (and a cure) is illustrated in detail in Dixon Chapters 5 and 9 [4] and in Dixon & Boone [12] and in the AAPM TG-111 report [11]. To wit, A pencil chamber cannot be used to directly measure the peak central dose $f(0)$, nor can the value of $f(0)$ be deduced (or even approximated) using a pencil chamber reading (even one of extended length), since such a reading represents the integral of $f(z)$.

Since $f(0)$ is the "point dose" on the central ray of the cone beam at depth in the phantom, the most obvious (and simplest) method is to directly measure the dose $f(0)$ at that point using a small ionization chamber (such as a 0.6 cc Farmer-type chamber) – the same method used for decades to measure depth- dose in a stationary phantom.

A study in simplicity compared to the integral method required for the CTDI paradigm which applies only to phantom-in-motion techniques – no pencil chamber required (or desired) in this case.

C. *Wide beam widths*. Another such IEC patch is a response to a paper by John Boone [16] which illustrates a significant drop-off in the value of $CTDI_{100}$ as the primary beam width becomes comparable to the pencil chamber length (nT > 40 mm). This patch is designed to keep $CTDI_{100}$ at the same fraction of $CTDI_\infty$ as that for narrow beams (this fraction being about 0.6 on the central axis of the body phantom). It does so for "phantom-in-motion" scan protocols, but it fails in the realm of stationary phantom dosimetry for which wide cone beams are more commonly used, and for which we provide the *appropriate correction* as shown in Chapter 9 [4] and in [12].

There is, inexplicably, no patch which provides a correction of $CTDI_{100}$ (thence $CTDI_{vol}$) for scan length using $CTDI_L = H(L)\,CTDI_{100}$ although a plethora of such robust $H(L)$ data exists as described in book Chapter 9 [4] as well as in other chapters. This correction would provide an appropriate (albeit approximate) physical interpretation for CTDI (TCM) as illustrated in Dixon's book [4], and in which rigorous methods of correcting $CTDI_{vol}$ for all modalities are provided.

XIII. USE OF THE SCANNER-REPORTED CTDI

Despite these differences, CTDI has been widely interpreted and used as an indicator of clinical patient dose by *regulators* and *medical physicists* alike, in *national dose surveys*, in *imaging literature*, *in the clinic*, etc.; and *on the CT monitor for every patient scan*.

XIV. SIZE-SPECIFIC DOSE ESTIMATES (SSDE)

The basic *SSDE* dose index concept presented in the Report of AAPM Task Group 204 [17] and as revised in [18] is an approach to develop a more reasonable *estimate* of patient dose using the scanner-reported $CTDI_{vol}$ and conversion factors that account for differing patient "sizes". In situations where a fixed tube current is employed and the patient anatomy and circumference is reasonably homogeneous over an entire CT scan, *SSDE* provides an improved *estimate* of dose as compared to $CTDI_{vol}$. The IEC has developed (but not yet implemented) a model by means of which SSDE will additionally be reported by the scanner which is based on a water-equivalent patient diameter *d* [18] , and once again using $CTDI_{vol}$ as a basis (and which SSDE values may soon be coming to a CT scanner near you). The various CT manufacturers will be responsible for the methodology (and validation of) the computation of water-equivalent diameter *d*, and thence SSDE.

XV. ESTIMATION OF ORGAN DOSES

There is a growing movement to calculate individual organ doses in CT, primarily based on Monte Carlo simulations, which begs the question: What are we to do with such data? Even if we could calculate organ doses accurately, are the risk factors for the individual organs that well known? Or will they even be?

Some *commercial dose-tracking software* now include an organ-dose computation for each patient; for example, by matching the patient's body habitus to a particular humanoid phantom on which Monte Carlo calculations of organ dose have been made. If these are further normalized to the patient, based on the scanner-reported value of $CTDI_{vol}$, then the above-mentioned caveats concerning $CTDI_{vol}$ remain in play.

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