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CRITERIA FOR ESTABLISHING SHIELDING OF MULTI-DETECTOR COMPUTED TOMOGRAPHY (MDCT) ROOMS

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The aim of this work is to compare two methods used for determining the proper shielding of computed tomography (CT) rooms while considering recent technological advances in CT scanners. The approaches of the German Institute for Standardisation and the US National Council on Radiation Protection and Measurements were compared and a series of radiation measurements were performed in several CT rooms at the Lausanne University Hospital. The following three-step procedure is proposed for assuring sufficient shielding of rooms hosting new CT units with spiral mode acquisition and various X-ray beam collimation widths: (1) calculate the ambient equivalent dose for a representative average weekly dose length product at the position where shielding is required; (2) from the maximum permissible weekly dose at the location of interest, calculate the transmission factor *F* that must be taken to ensure proper shielding and (3) convert the transmission factor into a thickness of lead shielding. A similar approach could be adopted to use when designing shielding for fluoroscopy rooms, where the basic quantity would be the dose area product instead of the load of current (milliampere-minute).

INTRODUCTION

The problem of scattered radiation around modern multi-detector computed tomography (CT) units is of current concern. On the one hand, the literature[1–4] reports that technological changes have introduced a modification of certain X-ray tube parameters that are sometimes used to calculate the shielding of a CT room. On the other hand, the number of CT examinations that can be performed during a day on a single unit has drastically increased over time. This means it has become necessary to verify whether the shielding methodology in use since CT was first introduced remains valid.

When designing the proper shielding of a CT room, the accurate determination of the spatial distribution of scattered radiation is necessary and various models have been proposed in the literature for this purpose[5–9]. Of particular interest are the models proposed by the US National Council on Radiation Protection and Measurements (NCRP)[10,11] and the German Institute for Standardisation (DIN)[12]. These models are the basis for shielding calculations in many countries around the world. Many works have been dedicated to the comparison of the various models[13–16] or to the comparison of the data obtained by these models with that provided by CT manufacturers[4,17,18].

In Switzerland today, the shielding of rooms with CT scanners is designed according to the Ordinance on Medical X-ray Units[19], which uses the same model as the one proposed by the DIN. It indicates the weekly tube loading needed for a CT slice, expressed in (milliampere-minute per week), and gives sets of minimal loading values depending on the size of the hospital.

The aim of this work is to compare the DIN model (the method used in Switzerland) and the NCRP model, to investigate the limits of using the tube loading in milliampere-minute as the working parameter for shielding new CT rooms and to explore the appropriateness of using the dose length product (DLP) instead.

METHOD

The NCRP concept

In the NRCP model, the air kerma is used to calculate secondary radiation (scattered and transmitted), whose contribution at 1 m from the isocentre is proportional to the DLP when scanning two CT dose index (CTDI) phantoms (trunk phantom: Ø32 cm and head phantom: Ø16):

$$K_{\text{trunk}} = \kappa_{\text{trunk}}\text{DLP and } K_{\text{head}} = \kappa_{\text{head}}\text{DLP}.$$

The κ factors given in publication NCRP 147, based on measurements and accounting for both scattered and transmitted radiation are

$$\kappa_{\text{trunk}} = 3.6 \times 10^{-4}\,\text{cm}^{-1} \quad \text{and}$$
$$\kappa_{\text{head}} = 0.9 \times 10^{-4}\,\text{cm}^{-1}.$$

The NRCP recommends using the dose line integral (DLI) instead of the DLP. The DLI is obtained by multiplying the DLP by a factor that varies with beam collimation and which is equal to 1.2 for the CTDI trunk phantom when the collimation is 20 mm[11]. This methodology makes it possible to include the part of scattered radiation produced in actual patients that is not recorded by the methodology of DLP measurements.

In this study, the air kerma established according to the NCRP method was converted into ambient dose equivalent $H^*(10)$ as follows:

$$H^*(10) = \theta.\,K \quad \text{with}\ \theta = 1.5\ \text{Sv Gy}^{-1}. \tag{1}$$

The German Institute for Standardisation concept

According to the DIN standard, calculating the equivalent dose of primary radiation ('Nutzstrahl'—H_N) is performed as follows:

$$H_\text{N} = WH_{\text{A},1}\left(\frac{1}{x^2}\right),$$

where W is the workload in milliampere-minute per week, $H_{\text{A},1}$ is the equivalent dose at 1 m per unit current loading in millisievert per milliampere-minute and x is the distance between the focal spot and the measurement point.

For example, in large hospitals, the minimal workload for a CT unit given in the DIN standard is 20 000 mA min week^{-1}, and the equivalent dose at 1 m per unit current loading given for 120 kV and 2.5 mm Al is 13 mSv (mA min)$^{-1}$, that is, 0.22 mSv (mA s)$^{-1}$.

Calculating the equivalent dose of scattered radiation ('Streustrahlung'—H_S) is performed as follows:

$$H_\text{S} = WH_{\text{A},1}\left(\frac{1}{a^2}\right)f_\text{K}\left(\frac{1}{s^2}\right)f_\text{D},$$

where a is the distance between the focal spot and the centre of the scattering phantom, s is the distance between the centre of the scattering phantom and the measurement point, f_K is the scatter yield ($f_\text{K} = 1.10^{-4}$ for CT).

Moreover, according to the DIN standard, the radiation transmitted across the tube housing is taken into account by multiplying the f_K factor by a coefficient f_D equal to three for CT.

$$H_\text{S} = WH_{\text{A},1}\left(\frac{1}{a^2}\right)f_\text{K}\left(\frac{1}{s^2}\right)f_\text{D}.$$

According to this relationship, the equivalent dose of scattered radiation, $H_{\text{S,trunk}}$, of the CTDI trunk phantom is $H_{\text{S,trunk}} = 26\ \mu\text{Sv}$ for 100 mA s at 120 kV and for a beam collimation of 10 mm.

This quantity can be expressed in terms of DLP, which is related to the tube loading (Q) through the normalised weighted CTDI (${}_\text{n}\text{CTDI}_\text{w}$ in mGy (mA s^{-1})) and the beam length (L):

$$\text{DLP} = Q\ {}_\text{n}\text{CTDI}_\text{w}\ L.$$

The dose of scattered radiation, H_S, is thus expressed as

$$H_\text{S} = \frac{\text{DLP}}{({}_\text{n}\text{CTDI}_\text{w}\ L)}\ H_{\text{A},1}\left(\frac{1}{a^2}\right)f_\text{K}\left(\frac{1}{s^2}\right)f_\text{D}.$$

Assuming a beam collimation of 10 mm, a tube voltage of 120 kV and considering a generic CTDI, it follows that

$$H_{\text{S,trunk}}(\text{mSv}) = 2.36 \times 10^{-3}\text{DLP (mGy cm)}. \tag{2}$$

For the head, the CTDI has to be divided by a factor of 2, and the same holds for the f_K factor.

Measurements

Dose measurements were performed on a 64-detector row CT system (VCT, GEMS, Milwaukee, USA) at the Lausanne University Hospital (CHUV). The CTDI trunk and head phantoms were scanned in the helical mode. The ambient dose equivalent, $H^*(10)$, was measured at various distances from the isocentre of the CT unit at various angles from the patient support to establish an isodose cartography. The measurements were performed using a Smartion dosemeter (Mini Instruments Inc., Burnham on Crouch, UK) calibrated in terms of $H^*(10)$ using a ^{137}Cs beam. According to the instrument sheet, a correction factor of 0.9 was applied to the measured values to take into account the difference in beam quality between ^{137}Cs and the CT scattered radiation. The collimation widths used during the acquisitions were 20 mm for the 32 × 0.625 configuration and 40 mm for the 64 × 0.625 configuration using 120 and 140 kV at various tube loadings. CTDI$_\text{vol}$ values indicated by the units were verified using a Radcal electrometer (Radcal Corp., Monrovia, CA, USA) connected to a standard CT ion pencil beam

Table 1. Scattered radiation per unit DLP established according to the NCRP and DIN models.

Model	H_s/DLP (μSv (100 mGy cm))$^{-1}$	
	Trunk	Head
NCRP	54	13.5
DIN	283–346	71–87

chamber. All dosemeters used in this study are traceable to the British National Metrology Institute (NPL).

Following this experiment, 10 sets of thermoluminescent dosemeters (LiF 100 TLDs) were placed for 2 weeks at various positions in the three rooms where a CT is present (two CT units used for elective examinations (a 64-detector row and an 8-detector row CT system; respectively, Lightspeed and VCT from GEMS), and one CT used for emergency situations running 24 h a day, a 64-detector row CT system (VCT from GEMS) in order to measure the ambient dose delivered in a 2-week period representing the normal use of the CT units.

RESULTS AND DISCUSSION

The $CTDI_{vol}$ is indicated by the unit and the measured values agreed within 5 %. Thus, all the DLP values used in this study are the ones given by the CT units. The monitoring of the data provided by the CT units (single- and multi-detector row CT) shows that the following conversion can be adopted: 5.0 ± 0.5 mGy cm (mA min)$^{-1}$. Using Equations (1) and (2), the dose of scattered radiation established by the NCRP and the DIN methods were calculated at a distance of 1 m for the CTDI trunk and head phantoms. The results are summarised in Table 1.

Table 1 shows that the ratio of the contribution of secondary radiation for the trunk and the head is 4; this corresponds to the ratio of the total energies imparted to the phantom. It is interesting to note that the NCRP model differs from the DIN model by a factor of about 5. However, there are two reasons that explain this difference. The first is that the DIN model considers an X-ray tube that has a total filtration of 2.5 mm Al in spite of the fact that CT units generally have a total filtration closer to 5 mm Al. The use of a more representative total filtration would have reduced the primary dose (and thus the secondary dose) by a factor close to 2. The second is that introducing a factor of 3 (F_D) concerning the transmission of radiation across the tube housing.

Figure 1 presents the dose of the scattered radiation at an angle of 45° from the patient support for 100 mA s and for the CTDI trunk phantom at various distances from the isocentre of the CT unit for two tube voltages (120 and 140 kV). As expected, the ambient dose equivalent $H^*(10)$ varies with the inverse of the square of the distance. It indicates also that at a distance of 1 m from the isocentre, for a tube voltage of 140 kV and a tube loading of 100 mA s, $H^*(10)$ equals 19.3 and 9.7 μSv for a 40 and 20 mm beam collimation, respectively. This result confirms that scattered radiation is proportional to the value of the beam collimation used.

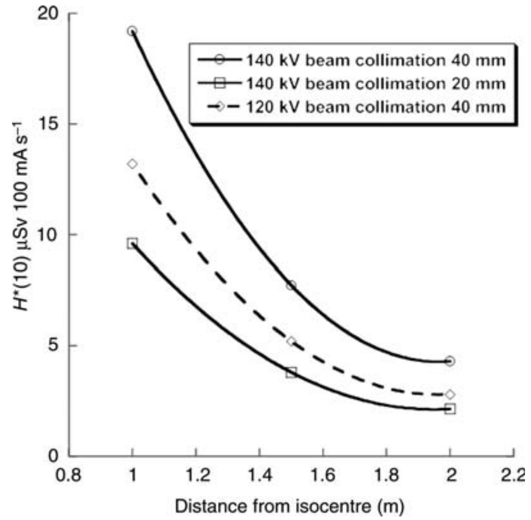

Figure 1. Dose of the scattered radiation per 100 mA s, at an angle of 45° from the patient support and for a CTDI trunk phantom at 120 and 140 kV.

Figure 1 also shows that for the same beam collimation, the tube voltage change from 120 to 140 kV is reflected by an increase of the scattered radiation by about 50 %. This increase is in agreement with the increase of the $_nCTDI_w$ when switching from 120 to 140 kV. This variation of the fraction of the scattered radiation for the same tube loading confirms the inappropriateness of using only tube loading to estimate the dose of scattered radiation around a CT unit.

According to the DIN approach, the expected value of scattered radiation at 1 m from the isocentre of the unit for a beam collimation of 10 mm is $H_s =$ 26 μSv (100 mA s)$^{-1}$ at 120 kV. The measurements show that for a beam collimation of 40 mm (condition used in Figure 1), the scattered radiation was only equal to 13 μSv (100 mA s)$^{-1}$; that is, half of what was predicted by the DIN standard. The use of a beam collimation of 10 mm would lead to one-fourth of this, roughly 3 μSv (100 mA s)$^{-1}$, and lead to an overestimation by a factor of 9.

Figure 2 presents the same data as in Figure 1 for the CTDI trunk and head phantoms but expressed in terms of DLP instead of tube loading (per 100 mGy cm rather than per 100 mA s).

It is shown that for a given diameter of the scanned object, the dose of scattered radiation does not depend on the beam collimation and registers a very weak dependence on tube voltage (3 % difference between 120 and 140 kV). The figure also indicates that the dose of scattered radiation, at 1 m for the CTDI trunk phantom is 31 μSv $(100\ \mathrm{mGy\ cm})^{-1}$ (54 μSv $(100\ \mathrm{mGy\ cm})^{-1}$ with the NCRP model and 240 μSv $(100\ \mathrm{mGy\ cm})^{-1}$ with the DIN model); for the CTDI head phantom it equals to 10.5 μSv $(100\ \mathrm{mGy\ cm})^{-1}$ (13.5 μSv $(100\ \mathrm{mGy\ cm})^{-1}$ with the NCRP model and 60 μSv $(100\ \mathrm{mGy\ cm})^{-1}$ with the DIN model). A difference of a factor of 3, instead of 4 as indicated above, is found between the CTDI trunk and head phantoms. This could be explained by the partial re-absorption of the scattered radiation within the trunk phantom.

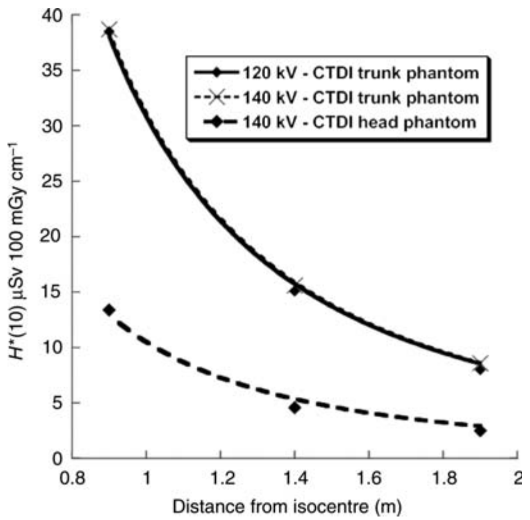

Figure 2. Dose of scattered radiation per 100 mGy cm^{-1}, at a 45° angle from the patient support for CTDI trunk and head phantoms.

In this case, the difference between the expected scattered radiation dose (54 μSv $(100\ \mathrm{mGy\ cm})^{-1}$) and the measured value (32 μSv $(100\ \mathrm{mGy\ cm})^{-1}$) shows that the NCRP approach leads, for this CT unit, to an overestimation lower than a factor of 2 as opposed to the DIN methodology that leads to an overestimation of a factor of 9.

Figure 3 presents the isodose curves in terms of the $H^*(10)$ of the scattered radiation obtained at a tube voltage of 120 kV around the CTDI trunk phantom. The associated $H^*(10)$ dose profiles as a function of distance both along the axis of the scanner and at a 45° angle are presented in Figure 4. The measurements were performed every 30 cm. Figures 5 and 6 present the same data as Figures 3 and 4, respectively, but for the CTDI head phantom.

The $H^*(10)$ dose profiles as a function of distance shown in Figures 4 and 6 vary slightly with the direction of the measurements (along the axis of the scanner and at a 45° angle). As expected, the scatter

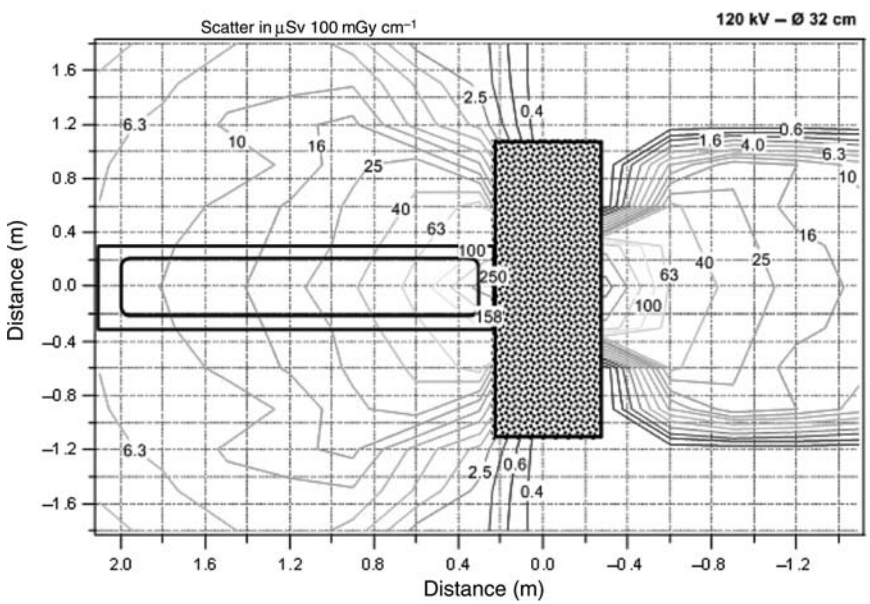

Figure 3. $H^*(10)$ isodose curves obtained at a tube voltage of 120 kV with a CTDI trunk phantom.

dose decreases as the inverse square distance from the isocentre of the unit. In Figures 4 and 6, the results obtained using the NCRP and DIN methods for a distance of 1 m are also represented for comparison. These figures show that the NCRP model leads to realistic values.

Figure 7 compares the expected values of the scatter dose according to the data presented in Figures 4 and 6 with the actual TLD measurements. The results show that no TLD measurement is over the predicted scatter dose value given by the NCRP approach for the CTDI trunk phantom. As expected, most of the measurements are within the curves obtained when scanning a CTDI trunk or head phantom. A few particularly low values were obtained when the TLDs were placed perpendicularly to the axis of the gantry at the level of the X-ray tube. These low values can be explained by the fact that the gantry itself offers some shielding.

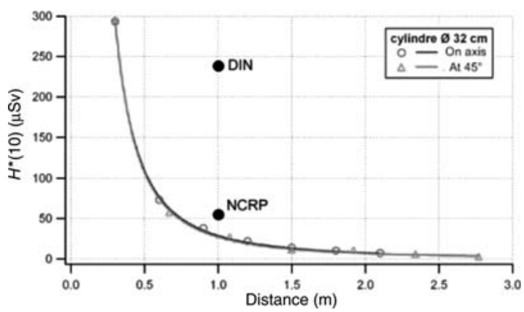

Figure 4. $H^*(10)$ as a function of distance along the axis of the scanner for a CTDI trunk phantom at 120 kV.

Table 2 presents the weekly milliampere-minute, the number of examinations, the DLP (mGy cm^{-1}) and the $H^*(10)$ in millisievert at various CT rooms. It is interesting to note that, for room 1, the change from a single-detector row to a multi-detector row CT led to an increase in milliampere-minute of 43 %, whereas this led only to a 23 % increase in DLP. Thus, the use of the DIN approach to shield room 1 would have overestimated the shielding by a factor of 2 in relative numbers. In fact, the actual values show that the DIN approach leads to an overestimation by a factor of 9.

According to these results, the following three-step procedure could be proposed for estimating the shielding of a CT room:

Step (1): Calculate the ambient dose equivalent, $H^*(10)$, for a given average weekly DLP (mGy cm^{-1}) and a given distance to the isocentre, assuming that all CT procedures are related to the trunk region (conservative approach). The $H^*(10)$ per week in mSv is expressed as the product of κ and DLP per week, κ being equal to 0.5 µSv (mGy cm)$^{-1}$ for CT examinations of the trunk. This approach is based on the NCRP philosophy.

Step (2): Calculate H_{lim} for the maximum permissible weekly dose at the location of interest, the transmission factor F that must be assured by the

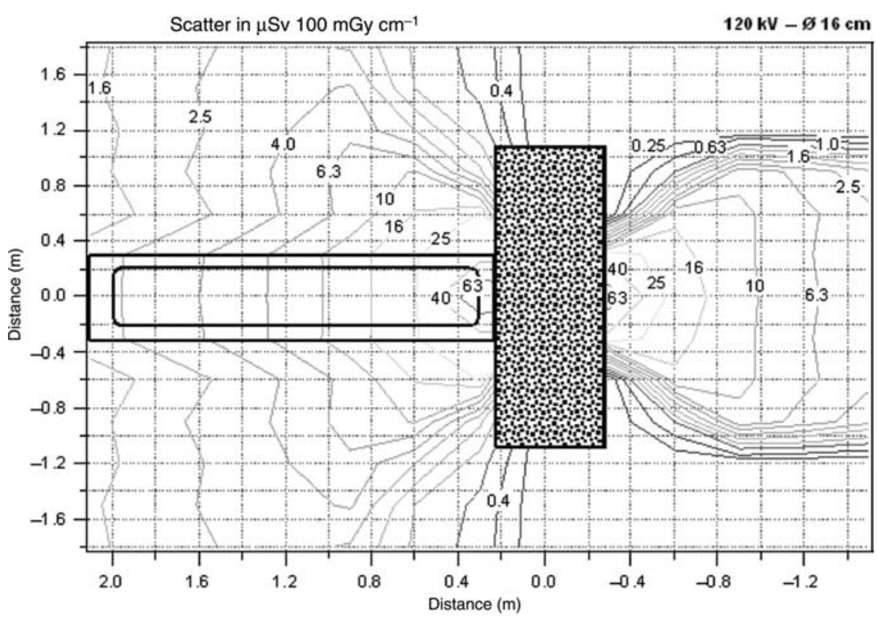

Figure 5. $H^*(10)$ isodose curves obtained at a tube voltage of 120 kV with a CTDI head phantom.

shielding. F is established as the ratio $H^*(10)$ over H_{lim}, H_{lim} being equal to 0.02 mSv week^{-1} or 0.1 mSv week^{-1}.

Step (3): Convert the transmission factor into a thickness of lead shielding, using Table A4 of the DIN standard 6812:2006-02.

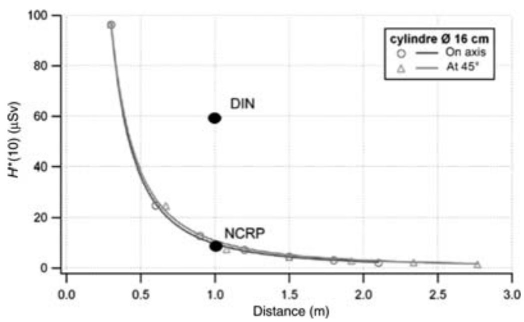

Figure 6. $H^*(10)$ as a function of distance along the axis of the scanner for a CTDI head phantom at 120 kV.

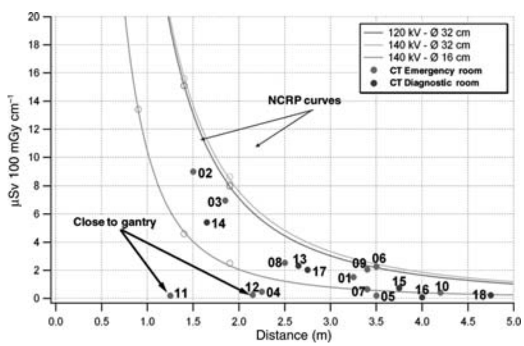

Figure 7. Comparison between the values established by interpolation from the isodoses shown in Figures 4 and 6 (expressed in $H^*(10)$) and the TLD measurements (expressed in $H_p(10)$).

A set of minimum DLP values could be proposed to ensure a reasonable shielding level in centres where the workload is expected to be particularly low. According to Table 2, the following values could be proposed: (i) radiology practice or small hospital: 50 Gy cm^{-1}; (ii) large or university hospitals where elective studies are performed: 100 Gy cm^{-1} and (iii) emergency unit: 200 Gy cm^{-1}.

CONCLUSION

In Switzerland, the tube loading needed for a CT slice, expressed in milliampere-minute, for a given anatomical region to be examined is used to establish the dose of scattered radiation and thus to design the necessary shielding of a CT installation. If this method was already questionable for single-detector row CT with sequential scanning (inadequate beam quality and tube leaking values), it becomes even more problematic with the introduction of spiral mode acquisition, because after a volume acquisition, as many slices as desired can be reconstructed.

This study showed that the rooms investigated were sufficiently shielded and certainly even highly over-shielded since the DIN approach was adopted to establish the shielding requirements. Although this overshielding means better protection for professionals or members of the public, it drastically increases the cost of the CT unit installation. To tackle this problem, the present DIN standard introduces the normalisation of the tube load by the beam collimation. However, the use of the tube load, which is not directly proportional to the amount of scattered radiation, is a solution that should be replaced by an approach that uses a quantity more closely linked to scattered radiation such as the DLP, even if the use of the CTDI and thus the DLP is questioned.

With the steady increase of the X-ray beam collimation width in modern CT units, the current

Table 2. Weekly milliampere-minute, number of examinations, DLP (mGy cm^{-1}) and $H^*(10)$ in mSv at various rooms/CT scanners at the Lausanne University Hospital (CHUV).

Room/CT scanner	mA min^{-1}	DLP trunk (number of examinations)	DLP head (number of examinations)	DLP total (number of examinations)	$H^*(10)$
Room 1: diagnostic CT (single-detector 1995)	9110		—	44 000 (est.) (75)	142 (DIN) 24 (NCRP)
Room 1: diagnostic CT (8-slice system 2008)	13 000	48 000 (95)	6000 (6)	54 000 (101)	203 (DIN) 27 (NCRP)
Room 2: diagnostic CT (64-slice system 2008)	n.a.	80 000 (106)	38 000 (25)	118 000 (131)	— (DIN) 47.4 (NCRP)
Room 3: emergency (64-slice system 2008)	n.a.	115 000 (175)	217 000 (138)	332 000 (313)	— (DIN) 79.9 (NCRP)

Est, estimated from protocols; n.a, not available anymore by the service engineer.

method needs to be replaced with a more robust one in order to assure sufficient shielding. The DLP should be used since it leads to results independent of the collimation and the tube voltage, and which depend only on the size of the scanned object.

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