TOPICAL REVIEW

Dose in x-ray computed tomography

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Topical Review

Dose in x-ray computed tomography

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Abstract
Radiation dose in x-ray computed tomography (CT) has become a topic of high interest due to the increasing numbers of CT examinations performed worldwide. This review aims to present an overview of current concepts for both scanner output metrics and for patient dosimetry and will comment on their strengths and weaknesses. Controversial issues such as the appropriateness of the CT dose index (CTDI) are discussed in detail. A review of approaches to patient dose assessment presently in practice, of the dose levels encountered and options for further dose optimization are also given and discussed. Patient dose assessment remains a topic for further improvement and for international consensus. All approaches presently in use are based on Monte Carlo (MC) simulations. Estimates for effective dose are established, but they are crude and not patient-specific; organ dose estimates are rarely available. Patient- and organ-specific dose estimates can be provided with adequate accuracy and independent of CTDI phantom measurements by fast MC simulations. Such information, in particular on 3D dose distributions, is important and helpful in optimization efforts. Dose optimization has been performed very successfully in recent years and even resulted in applications with effective dose values of below 1 mSv. In general, a trend towards lower dose values based on technical innovations has to be acknowledged. Effective dose values are down to clearly below 10 mSv on average, and there are a number of applications such as cardiac and pediatric CT which are performed routinely below 1 mSv on modern equipment.

1. Introduction
Dose in x-ray computed tomography (CT) has received appropriate attention since the introduction of CT in the 1970s; efforts addressed technical and physics issues and covered both scanner characteristics and patient dose aspects. Dose metrics for characterizing CT
Figure 1. Although CT examinations make up only a relatively small part of the total number of x-ray examinations, CT contributes a high proportion of the collective effective dose. The figures shown refer to Germany for the year 2006 but can be considered representative (BFS2009).
Figure 2. 2D projection radiography often suffers from details being obscured by overlaying anatomical structures (a). 3D slice imaging reliably removes such effects as is shown for a solitary nodule (see arrow) in this exemplary case (b) (images: courtesy of M Lell, University of Erlangen).

even very low doses can be detrimental or lethal although this is not supported by scientific evidence (Feinendegen et al 1999, Tubiana et al 2009, Hendee and O’Connor 2012, Neumaier et al 2012). Risk issues will be commented on further below after analyzing the dose situation and quantifying dose levels.

The reference to high cumulative doses due to CT practice automatically tends to link CT to high dose and risk per patient exam in the perception of the general public. In consequence, CT received increased scrutiny with respect to dose since about the start of the new millennium. This can be considered positive since it led to a number of valuable initiatives worldwide aiming for dose surveys and reviews of CT practice, for example in the USA the campaigns ‘Image Gently’ and ‘Image Wisely’ (www.imagegently.org; www.imagewisely.org). The situation also triggered a number of effective technical developments and fruitful initiatives leading to a substantial reduction of the dose to the patient per exam. Currently, low-dose CT and even ‘sub-mSv CT’ exams have become realistic options (Kalender 2011, McCollough et al 2012) and are in clinical use.

In parallel to these technical developments, we also witnessed debates about scanner and patient dosimetry, primarily in the USA (Dixon et al 2005, Boone 2007, AAPM 2010). They resulted in a number of new proposals which are still under discussion. This review aims to address all the topics above and to point to possible or probable solutions. Approaches to estimate patient dose are of particular concern. Concepts that provide scanner-, scan parameter-specific, patient- and organ-specific dose estimates (POSDE) are of particular interest. New scanner types such as flat-detector cone-beam CT, breast CT and other dedicated designs which are already in use in radiological practice in increasing numbers are to be included in these considerations.

2. Technical measures of the dose output of CT scanners

2.1. The CT dose index

CTDI concepts are well documented and have recently been re-evaluated (AAPM 2010, IEC 2010, IAEA 2011, Kalender 2011); only a brief summary is intended here. The CTDI is a measure of the CT scanner’s radiation output normalized to a tube current time product, usually quoted in units of mGy/100 mAs. It aims at determining an exposure rate characteristic for a
volume exam which, due to scattered radiation contributions from the surrounding media, is always higher than the dose measured for a single-slice exposure. This effect is sketched in figure 3 (Kalender 2011).

The concept of the CTDI was proposed and established in the USA by the Food and Drug Administration (FDA) in the late 1970s (Shope et al 1981). It is measured in polymethyl methacrylate cylinders of 15 cm in length and diameters of 16 and 32 cm considered representative for a standard head and body section, respectively, with one central and four peripheral bore holes in which a 100 mm long ionization chamber is inserted (figure 4). It is determined as

$$CTDI_{100,x} = \frac{1}{N \cdot T} \cdot \int_{-50\text{mm}}^{+50\text{mm}} D(z) \, dz,$$

(1)

$T$ is the section thickness, $D(z)$ the dose distribution along the $z$-axis, $N$ refers to the number of sections or slices measured simultaneously. The subscript $x$ refers to the measurement position which can be central (c), peripheral (p) or in air (central in the field of measurement but without a phantom). To take the inhomogeneity of dose distributions into account, a weighted CTDI was agreed upon

$$CTDI_w = \frac{1}{3} \cdot CTDI_{100,c} + \frac{2}{3} \cdot CTDI_{100,p}.$$  

(2)
Up to around 2005 the CTDI concept was not questioned. It was the standard for acceptance and constancy testing of CT apparatus and served its purposes well for scanner and protocol comparisons. New challenges arose when wider CT detectors were introduced with collimation widths nearing or exceeding the 100 mm length of the standard CT ionization chamber. In such a case, obviously, primary radiation is not registered completely and multiple scatter contributions only to a limited extent. The situation was first addressed by Mori et al (2005) and by Boone (2007). Boone showed that CTDI values measured in a 15 cm phantom with a 10 cm ionization chamber are lower than for a phantom infinite in length in any case. The ratio of CTDI_{100} to the limiting value for infinite phantom length was termed CTDI efficiency; it was found to be nearly constant up to 40 mm collimation but falls off significantly for wider collimation. 100% efficiency is not considered a necessity because this cannot be expected in patient scanning either. More or less constant CTDI_{100} values for varying collimation widths at otherwise constant parameters are desirable, however.

Two solutions were offered for wide collimations. The AAPM Task Group 111 considered in AAPM Report 111 (AAPM 2010) to use extended phantoms, e.g. a single 50 cm long phantom or three phantoms each 30 cm in length lined up in row. Measurements were proposed using a small volume ionization chamber with table translation. Final prescriptions on how to carry out such measurements are still in preparation.

In parallel, an official regulation was issued by the International Electrotechnical Commission (IEC) in 2010 (IEC 2010) which confirmed the validity of the existing CTDI regulations for collimation widths up to 40 mm and introduced a correction factor based on the ratio of two CTDI measurements in air with a 300 mm ionization chamber for scanners with wider collimations

\[
\frac{\text{CTDI}_{100}}{(N \cdot T)_{\text{ref}}} = \int_{-50\text{mm}}^{+50\text{mm}} D_{\text{ref}}(z) \, dz \cdot \frac{\text{CTDI}_{\text{free in air}, \text{ref}}}{\text{CTDI}_{\text{free in air}, N \cdot T}}.
\] (3)

The reference collimation ‘ref’ is to be chosen at 40 mm or below; IEC thereby aims to provide reliable CTDI values independent of the collimation width. The proposal meanwhile was set into effect, it appears reasonable and above all practical. It relies on the established CTDI phantoms and only requires an additional 300 mm long ionization chamber and one additional measurement if CTDI values for a scanner with wide detector are required. It is of interest to note that this solution can also be adapted to provide regulations for flat-detector CT which also can have collimation widths beyond 40 mm (Kalender 2011).

2.2. Technical descriptors of exam-specific dose levels

Dose estimates are intended to provide values in mGy. For this end, the applicable normalized CTDI value is multiplied by the mAs product Q used in the exam. Division by the pitch factor \( p \) allows taking spiral CT acquisitions into account appropriately and yields the so-called ‘volume CTDI’ or CTDI_{vol}:

\[
\text{CTDI}_{\text{vol}} = \text{CTDI}_0 \cdot \frac{Q}{p[mGy]}.
\] (4)

A quantity equivalent to CTDI_{vol} can be provided for sequential CT, i.e. for series of single section scans, in an analogous manner. In the standard case of contiguous gapless scans \( p = 1 \) applies directly to equation (4); differences in spacing of scans can be taken into account by setting \( p \) accordingly.

Statements of CTDI_{vol} values are obligatory for each CT exam and are required to be provided by the manufacturers on the CT scanner console. However, CTDI_{vol} neither takes the patient size and cross-section into account nor the length of the scanned volume. More
information is provided by the dose-length product (DLP) which is obtained by multiplying the CTDI\text{vol} value by the scan range $R$:

$$DLP = \text{CTDI}_{\text{vol}} \times R [\text{mGy} \cdot \text{cm}].$$ \hfill (5)

The quantity DLP apparently does not provide a dose value as is also evident from its unit mGy \cdot cm. Nevertheless it is a useful quantity of interest which serves as a surrogate for patient dose. This is especially meaningful in efforts to compare dose levels, and it became accepted through the establishment of diagnostic reference levels (DRL) (EC 1999, Shrimpton 2004, AAPM 2008, Kalender 2011).

However, both CTDI\text{vol} in mGy and DLP in mGy \cdot cm only provide rough estimates of the patient dose levels involved. They are determined for cylindrical phantoms and do not match the patient situation well (Kalender 2011, McCollough et al. 2011). Patient-specific dose estimates which take patient size, cross-section and the anatomic scan range into account are preferable.

2.3. Summary

The CTDI concepts presently in use fulfil their tasks for scanner characterization, acceptance and constancy testing well and are established worldwide. The derived quantities CTDI\text{vol} and DLP which follow directly for each CT exam are very useful as is reflected in the concept of DRLs. These serve an important function in guiding users with respect to parameter selections. However, all the descriptors discussed in this section are based on standard CTDI phantoms only and do not take patient specifics into account. In consequence, they do not provide patient dose or organ dose values.

3. Estimates of patient dose

The technical dose descriptors discussed above are clearly defined; the measurement procedures are fixed and prescribed adequately in official regulations. The situation regarding estimates of patient dose is much more complex. This is mostly due to the large variation in patient size and anatomy. For a given set of exposure parameters, the dose in the center of the patient or phantom will vary strongly depending on the diameter, i.e. on the tissue and fat layers around it which attenuate the radiation intensity. For 20, 40 and 60 cm diameter water cylinders, for example, measured at 120 kV and 100 mAs in a typical CT scanner, dose values of 6.7, 1.3 and 0.2 mGy, respectively, result in the center. This reflects the well-known fact that fat layers can be protective to some degree with respect to dose. Apart from patient size there are other problems to be considered regarding the determination of ‘patient dose’, in particular for organ dose values and effective dose, as will be discussed below.

3.1. Estimates based on CTDI phantom measurements

A recent effort to take patient size into account was started by the AAPM Task Group 204 who proposed size-specific dose estimates (SSDE) (AAPM 2011). The goal was to adapt the CTDI\text{vol} provided on the CT scanner console for the patient exam in question to the patient anatomy. As indicated above by the dependence of dose levels on phantom diameters, dose for larger patients will be lower than the CTDI\text{vol} value measured in a 32 cm diameter phantom and higher for very slim patients. While the respective trends are predictable, numerical correction values are not easy to determine. As described in detail in AAPM Report no 204 (AAPM 2011) SSDE is to be estimated using the anthropometric parameter ‘patient diameter’. Tables
The family of ORNL phantoms provide adapted versions of ‘standard man’ to represent different age categories and genders here shown by coronal CT slices. All organs of interest with respect to radiation exposure are mathematically described allowing for objectively defined and reproducible dosimetry.

of conversion factors $f$ are provided in the report to adapt the standard measured CTDI value to

\[ \text{SSDE} = f \times \text{CTDI}_{\text{vol}}. \]

The extensive tabulations in AAPM Report 204 were derived by Monte Carlo (MC) calculations and fit procedures and allow selection of $f$ as a function of the chosen diameter and the CTDI phantom used for either the 16 cm or 32 cm diameter CTDI value. It is not trivial, however, to determine patient diameters. The report offers four different choices of patient diameter: lateral, anterior–posterior (a.p.), lateral + a.p., or effective patient diameter. Effort is required to ensure consistent use of the data but this may be achieved. More difficult is the objective definition of patient diameters no matter which one is chosen; they vary with the anatomic level and may change strongly over the examined range.

Further consensus decisions and instructions on determining patient diameters appear necessary for a general acceptance of the SSDE concept. Also, steps toward matching the present DLP-to-E concept (see 3.2.1) are still missing.

3.2. Estimates based on MC calculations using anthropomorphic phantoms and CTDI in-air measurements

Substantial efforts at developing methods for estimating patient dose in CT were initiated and funded in the 1990s in Europe by the European Commission. The approach generally chosen was based on MC calculations using anthropomorphic phantoms and CTDI measured in air for calibration purposes (Shrimpton et al 1991, Zankl et al 1991, EC 1999, Kalender et al 1999a). Any assumption that CTDI phantoms represent the patient situation was thus avoided. Thereby the resulting estimates of patient dose are independent of the technical exposure measures using CTDI phantoms which were discussed in section 2.

The family of anthropomorphic phantoms developed and provided by the Oak Ridge National Laboratories (ORNL) includes adults, children and newborns of both sexes (figure 5). The adult ‘standard man’ is the best known example and represents an average person of...
Table 1. Factors for the conversion of DLP to \(E\). Values for adults. (EC 1999, Shrimpton 2004, Deak et al 2010).

<table>
<thead>
<tr>
<th>Body region</th>
<th>All ages (EC 1999)</th>
<th>Adult (Shrimpton 2004)</th>
<th>Adult (Deak et al 2010)</th>
<th>Adult (\sigma) (Deak et al 2010)</th>
<th>Adult (\varphi) (Deak et al 2010)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
<td>0.0023</td>
<td>0.0021</td>
<td>0.0019</td>
<td>0.0018</td>
<td>0.0020</td>
</tr>
<tr>
<td>Neck</td>
<td>0.0054</td>
<td>0.0059</td>
<td>0.0052</td>
<td>0.005</td>
<td>0.0053</td>
</tr>
<tr>
<td>Thorax</td>
<td>0.017</td>
<td>0.014</td>
<td>0.015</td>
<td>0.011</td>
<td>0.019</td>
</tr>
<tr>
<td>Abdomen</td>
<td>0.015</td>
<td>0.015</td>
<td>0.015</td>
<td>0.013</td>
<td>0.017</td>
</tr>
<tr>
<td>Pelvis</td>
<td>0.019</td>
<td>0.015</td>
<td>0.013</td>
<td>0.01</td>
<td>0.016</td>
</tr>
</tbody>
</table>

170 cm body height and 75 kg weight. Organs of interest with respect to patient dosimetry are mathematically defined for all phantoms; respective representations as voxel phantoms are available and form the basis for MC calculations of 3D dose distributions. Organ dose values are directly available as the mean value of all voxels assigned to the respective organ. Effective dose calculated as the weighted sum of organ dose values as prescribed by ICRP directly results (ICRP 2007). This option has been used in several ways since the 1990s as shown in the next sections.

3.2.1. DLP-to-\(E\) conversion. MC calculations were first carried out for typical CT examinations of the head, chest, abdomen and pelvis of the adult phantoms using standard scan parameters and scan ranges. Since the DLP is defined for that situation and effective dose \(E\) results from MC calculations, the ratio of DLP-to-\(E\) is given for the respective anthropomorphic voxel phantom. This provided the basis for establishing DLP-to-\(E\) conversion coefficients \(k\) (EC 1999). This concept is limited to standard phantoms but it is very useful in giving a single number for orientation and is widely used internationally. It seems to be common practice in publications today to quote effective dose values; interestingly, the quantity \(E\) used to specify dose is not even specified explicitly in many cases, e.g. ‘coronary CT with a consistent dose below 1 mSv using prospectively electrocardiogram-triggered high-pitch spiral acquisition’ (Achenbach et al 2010).

DLP-to-\(E\) coefficients are widely available and have been extended to cover both sexes, children and newborns (Huda et al 2008, AAPM 2010, Deak et al 2010). Recently they have also been revisited and adapted to modern CT scanners and wide detectors, a range of voltage values, and all ages available in the ORNL family (Deak et al 2010). Some representative \(k\)-values and their changes over time are shown in table 1. They are officially endorsed and serve their purpose well.

Limitations of both the concept of effective dose \(E\) and of DLP-to-\(E\) conversion have to be stressed, however. The choice of ‘body region’ is not always easy and unambiguous, and each \(k\)-value is applied for all CT scanners alike which contributes to overall uncertainty. Estimates of \(E\) are phantom- and not patient-specific, they do not take patient size into account, and they do not provide organ dose values. In addition, it has to be noted that effective dose is not an appropriate measure for individual patients but only for comparison of technologies and patient groups (ICRP 2007).

3.2.2. Organ dose estimates based on pre-tabulated phantom data. The idea to provide organ dose values instead of, or in addition to, effective dose led to a number of efforts and software programs in Europe since the 1990s providing organ dose estimates for standard
The calculation of organ dose values and effective dose for teaching and training purposes is possible interactively for any arbitrary CT scanning protocol. The results are provided in graphical and in tabular form.

3.2.3. Patient- and organ-specific dose estimates based on MC calculations. A relatively new approach of POSDE (Kalender 2012a) is illustrated in figure 7 and proceeds in the following manner. (1) The patient CT data are used for providing full information about the patient’s shape in the scanned region; this avoids the user having to estimate representative a.p. and lateral diameters or other. (2) The patient’s CT data are appended at the top and at the bottom of the available image volume by data taken from size-adapted standard anthropomorphic voxel phantoms. The ORNL family of phantoms (Cristy and Eckermann 1987) represents an adequate choice, but alternatives are available and may be used in this concept in an analogous manner. In any case, the procedure results in a patient-specific whole-body voxel phantom suitable for MC dose calculations with improved modeling of single and multiple scatter.
Figure 7. Approach to patient- and organ-specific dose estimates (POSDE) which uses the patient CT data appended by data from size-adapted anthropomorphic voxel phantoms for generating patient-specific whole-body voxel phantoms for accurate MC dose calculations.

Feasibility of the concept of generating whole-body voxel phantoms as a combination of patient and phantom data is straightforward; voxel phantoms are completed rapidly but there is need for controlling organ size and position. Comparisons of dose values measured by TLD to those calculated by MC methods were performed on a point-by-point basis using an adult and two pediatric phantoms each loaded with TLD chips. These experiments yielded deviations of less than 10% on average when using whole-body voxel phantoms in MC calculations. When using only the available CT images and not the whole-body model, deviations increased strongly, in particular for the border regions. The excellent agreement between measured and calculated dose values was achieved because all necessary parameters of the CT scanner and scan were known. Accuracy of 10% or better may be desirable in research including technical and phantom measurements, but certainly is not mandatory for routine or clinical applications. Statistical errors in MC-based methods are not a problem because simulations can work with arbitrarily high photon numbers without affecting the patient.

All these statements refer to estimates of dose for organs which are clearly delineated. Distributed organs and tissues such as skin, mucosa and bone marrow are hard or impossible
to handle in practice, however. In consequence, estimates of effective dose are not possible in the standard way because dose cannot be determined for all organs required by ICRP for the calculation of effective dose (ICRP 2007). This is not considered a serious issue since ICRP also states that the concept of effective dose is not to be applied to individual patients. A practical example of POSDE is the estimation of embryo dose from abdominal/pelvic CT examinations performed on pregnant patients (Damilakis et al 2010a, 2010b). Similar to the situation with SSDE, POSDE work is still in progress and there are no official regulations.

3.3. Summary

Patient dose estimates in CT are primarily based on MC calculations using anthropomorphic phantoms because direct measurements in vivo are neither practical nor do they allow the assessment of the spatial distribution of dose. Estimates of effective dose E based on DLP-to-E conversion are in common use and serve their purpose of general dose surveillance well, but are not sufficient to indicate dose to individual patients. Patient-specific dose estimates such as SSDE and POSDE are still work in progress. The important message is that organ doses can be estimated with high accuracy; such efforts are demanding, and are neither indicated nor necessary for routine purposes.

4. Typical patient dose values in clinical CT exams and current trends

4.1. Development of patient dose values over the past decades

There have been numerous surveys on patient dose in CT in different countries and regions and with varying focus with respect to CT applications and anatomic ranges. A good general overview on long-term developments has been provided by UNSCEAR (table 2) (UNSCEAR 2010). Patient dose was relatively low in the 1970s, the first decade of clinical CT practice, which was essentially limited to CT of the brain. The low value of 1.3 mSv effective dose reflects the fact that both the scan range and the available x-ray power were rather limited initially. The increase in the 1980s reflects the introduction of whole-body CT which started in the late 1970s and the necessary increases in x-ray power. The trend continued into the 1990s which began with the introduction of spiral volumetric CT (Kalender et al 1990). The increase in effective doses to 8.8 mSv was not related to the spiral scanning approach itself but due to the increasingly larger scan volumes which became practical. Spiral CT offered considerable diagnostic advantages and new applications, as for example provided by the newly available spiral CT angiography which started in the early 1990s (Bautz et al 1991). The trend of increasing patient dose values continued into the 2000s since increased x-ray power and further clinical applications such as cardiac CT became available (Kachelrieß and Kalender 1998, Kalender 2006). At the same time, awareness of patient dose and dose reduction developments gained increasing importance in radiology. This stopped the trend toward increasing doses; UNSCEAR states mean effective dose values of 7.4 mSv for the early 2000s.

More detailed data regarding dose values for specific CT applications were provided by the European Community as shown in table 3 (European Commission 2008). These data are in
general agreement with the UNSCEAR data and confirm that the upward trend in dose values apparently has been reversed. The trend toward lowering patient doses appears to continue; although there are no survey data available yet for the 2010s, there are many indications and publications pointing in that direction.

### 4.2. Technological efforts at dose optimization

There always was consensus that dose should be kept as low as reasonably achievable (ALARA). The ALARA principle was originally defined for occupational exposure but is now routinely applied to CT and to any other application of ionizing radiation in medicine. This was initially done by focusing on traditional measures such as limiting the mAs value, considering higher pre-filtration, and strictly controlling the scan range. Such measures are a necessity and analogous to the practice in conventional radiography and are largely the task of the CT operator.

Starting in the late 1990s, the CT manufacturers were increasingly challenged to focus on dose issues. In consequence they stepped up their efforts to improve their scanners’ dose efficiency. This meant questioning the technical solutions established over the first three decades of CT practice and resulted in a number of smart and innovative technical solutions for dose optimization (table 4). These developments have been described fully in original publications; only short descriptions are given here.

**TCM and automatic exposure control (AEC).** Tube current in CT traditionally was held constant at pre-selected values for the duration of the scan. Until the 1990s, there were no efforts yet regarding TCM or an AEC although such concepts were already established in radiography and fluoroscopy. The advent of continuous data acquisition based on modern slip ring technology allowed not only for spiral scanning but inherently also for continuous

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**Table 3.** Mean effective doses for the CT exams surveyed in the ten European countries participating in DOSE DATAMED. (EC 2008).

<table>
<thead>
<tr>
<th>Exam type</th>
<th>LU</th>
<th>BE</th>
<th>DE</th>
<th>NO</th>
<th>CH</th>
<th>FR</th>
<th>SE</th>
<th>DK</th>
<th>NL</th>
<th>UK</th>
<th>Min</th>
<th>Max</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head</td>
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<td>2.3</td>
<td>2.6</td>
<td>1.8</td>
<td>2.2</td>
<td>1.8</td>
<td>2.0</td>
<td>1.9</td>
<td>1.2</td>
<td>2.0</td>
<td>2.2</td>
<td></td>
</tr>
<tr>
<td>Neck</td>
<td>2.5</td>
<td>–</td>
<td>2.5</td>
<td>3.4</td>
<td>3.1</td>
<td>2.5</td>
<td>–</td>
<td>1.3</td>
<td>–</td>
<td>2.4</td>
<td>2.6</td>
<td></td>
</tr>
<tr>
<td>Chest</td>
<td>10.0</td>
<td>4.1</td>
<td>7.6</td>
<td>11.5</td>
<td>8.8</td>
<td>5.5</td>
<td>–</td>
<td>11.0</td>
<td>5.5</td>
<td>7.8</td>
<td>2.8</td>
<td></td>
</tr>
<tr>
<td>Spine</td>
<td>9.0</td>
<td>–</td>
<td>2.9</td>
<td>4.3</td>
<td>9.1</td>
<td>4.0</td>
<td>–</td>
<td>5.7</td>
<td>3.1</td>
<td>4.2</td>
<td>3.1</td>
<td></td>
</tr>
<tr>
<td>Abdomen</td>
<td>15.0</td>
<td>11.3</td>
<td>18.6</td>
<td>12.6</td>
<td>8.4</td>
<td>5.8</td>
<td>–</td>
<td>14.0</td>
<td>10.6</td>
<td>9.8</td>
<td>3.2</td>
<td></td>
</tr>
<tr>
<td>Pelvis</td>
<td>–</td>
<td>–</td>
<td>10.6</td>
<td>9.3</td>
<td>7.0</td>
<td>–</td>
<td>8.3</td>
<td>7.4</td>
<td>9.8</td>
<td>1.5</td>
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<td>Trunk</td>
<td>7.9</td>
<td>–</td>
<td>24.4</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>10</td>
<td>15.0</td>
<td>–</td>
<td>10.4</td>
<td>3.1</td>
<td></td>
</tr>
<tr>
<td>Mean over all exam types</td>
<td>7.4</td>
<td>7.7</td>
<td>8.1</td>
<td>6.1</td>
<td>6.0</td>
<td>3.5</td>
<td>6.0</td>
<td>5.9</td>
<td>5.3</td>
<td>5.4</td>
<td>2.3</td>
<td></td>
</tr>
</tbody>
</table>

**Table 4.** Innovative developments in CT technology aiming at higher dose efficiency.

<table>
<thead>
<tr>
<th>Topics</th>
<th>Year of introduction</th>
<th>Dose reduction potential, %</th>
</tr>
</thead>
<tbody>
<tr>
<td>TCM</td>
<td>1999</td>
<td>10–60</td>
</tr>
<tr>
<td>Optimization of x-ray spectra</td>
<td>2009</td>
<td>10–50</td>
</tr>
<tr>
<td>X-ray beam collimation</td>
<td>2009</td>
<td>5–30</td>
</tr>
<tr>
<td>Iterative image reconstruction</td>
<td>2010</td>
<td>10–60</td>
</tr>
<tr>
<td>More efficient x-ray detectors</td>
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monitoring of attenuation values as a function of projection angle and position along the longitudinal axis. This opened the possibility of adapting the tube current to attenuation in real time during the scan. The necessary technical solutions for TCM were not trivial but were successfully implemented within a short time.

First clinical trials of TCM in the late 1990s showed convincing results (Kalender et al 1999b, 1999c). The principle is (1) start the scan with an adequate pre-selected tube current value which ensures that sufficient signal is recorded also for the most strongly attenuating projections, usually the lateral ones; (2) reduce the tube current wherever attenuation is lower. This leads to strongly reduced mAs values per rotation without a reduction of image quality as illustrated by the example in figure 8. Since attenuation in the a.p. direction is typically up to a factor of 10 lower than in the lateral direction in the shoulder region the reduction of the tube current by up to a factor of 10, i.e. by 90\%, which is technically feasible does not cause a visible increase of noise. An increase of the peak current value was used for the exam in figure 8(b) aiming at a reduction of noise in the lateral direction; the total mAs product for the scan shown was reduced by 49\% in spite of that increase when compared to the original scan.

It should be stated that dose reduction values are always higher than the mAs reduction values; this is due to the fact that dose to the central organs stems mostly from the a.p. projections which are less attenuated and thereby have higher intensities. Typical values of 50\% mAs reduction for the shoulder region amount to a dose reduction of about 60\% to 70\% (Kalender 2011). This indicates the high potential of dose reduction by modulation of the tube current with attenuation measured as a function of projection angle $\alpha$ in real time.

In addition to modulation with projection angle $\alpha$ (‘$\alpha$-modulation’) TCM technology also allows adapting the mAs value per rotation to changes in the patient cross-section along the longitudinal or z-direction (‘$z$-modulation’). This option can be used in a manner analogous to $\alpha$-modulation to adapt the user-selected mAs product per rotation whenever the cross-section and thereby also the x-ray attenuation decrease. A pragmatic approach proposed in 2001 (Kalender 2001) allows the user to select a noise level instead of an mAs value. The scanner then adapts the tube current continuously to deliver an approximately homogeneous noise level for the complete scan range; this and other AEC concepts are currently offered on all modern CT scanners. An example is illustrated in figure 9 by the coronal views of the 3D dose distributions obtained by MC calculations for scans without and with AEC activated. In this case noise homogeneity was vastly improved while at the same time reducing the mAs product by 34\% and dose by 45\%. TCM and AEC offer dose reduction potentials of a wide

Figure 8. Anatomy-adapted TCM. Since attenuation may vary strongly with projection angle, the constant tube current (a) can be increased slightly laterally and decreased strongly in the a.p. direction (b) resulting in this case in an mAs reduction of 49\% (reproduced with permission from Kalender 2001).
range between 10% and 60% depending mostly on the patient’s shape. For a perfect cylindrical shape, characteristic of a phantom but never of a patient, there is no modulation indicated.

**Optimization of x-ray spectra.** Similar to keeping the tube current constant, x-ray spectra were also initially kept constant. For the first four decades of CT practice, the 120 kV value chosen by G. Hounsfield in 1970 was predominantly used and provided by all CT manufacturers. There are good reasons for that choice: High x-ray intensity was a prerequisite for examinations with appropriately short scan times necessary to prevent patient motion artifacts. Since intensity increases roughly proportional to the square of the voltage, high kV values provide clear advantages; 140 kV, the highest value offered on standard equipment, is used as an alternative for obese patients, but associated with a decrease in contrast. Systematic investigations of the contrast-to-noise ratio obtainable per unit of dose as a function of beam energy and of the contrast mechanism involved were reported (Kalender et al 2007) and published (Kalender et al 2009) in the late 2000s. Based on simulations and on phantom measurements spanning a wide range of energies and patient cross-sections (figure 10(a)) it was shown that the optimal energy depends strongly on the type of contrast under consideration. For soft tissue or density differences, 120 kV appears to be a good choice (figure 10(b)). For skeletal and for contrast medium-enhanced imaging, i.e. for imaging of the elements calcium and iodine, much lower voltage values are indicated (figures 10(c) and (d)). The optimal spectra were found far below 80 kV, but are not practical due to the limited intensity and the high attenuation at low energy. But the realistic option of going from 120 to 80 kV offers dose reduction potentials of typically 30% to 50%.

The CT manufacturers reacted to these new findings and provided dedicated scan protocols within a very short time span. Cardiac and pediatric CT are primary fields of application for low
Figure 10. Results for spectral optimization in CT based on simulations and measurements for thoracic CT using anthropomorphic phantoms and a specimen for pediatric CT (a). There is only limited optimization potential for imaging of density differences (b), but a surprising potential for calcium and iodine imaging (c), (d) (reproduced with permission from Kalender 2011).

kV values because of the lower power requirements. In general, efforts at providing a broader range of voltage values both below 80 kV and above 140 kV, the first for contrast-enhanced, the latter for soft-tissue imaging of obese patients, are ongoing. The use of spectra optimized to the specific application which was neglected for a long time has started and may provide further improvements of image quality at reduced dose.

**X-ray beam collimation.** Issues with respect to collimation arose with the introduction of spiral CT with wide detectors. They are referred to as overscanning and overbeaming (Kalender
Iterative image reconstruction. Iterative reconstruction (IR) methods have recently re-emerged in transmission x-ray CT. They have been used in nuclear medicine for decades and were also used in CT in its very early years. Due to the much higher computational demands imposed by IR compared to analytical methods, they had to be abandoned when the amount of measured data acquired per scan was increased to allow for higher image quality. The increasing availability of affordable computational power has changed the situation; IR has become a hot topic in the past years and is aimed primarily at noise reduction, i.e. dose-efficient image reconstruction. CT manufacturers drove the development of IR approaches, and product announcements appeared in the late 2010s. All major CT manufacturers today provide IR software packages and continuously feature impressive image results; a clinical example is shown in figure 11.

There is no doubt that a large dose reduction potential has been provided by IR because images at a given acceptable noise level can be acquired with lower dose when using IR methods. The USA Federal Drug Administration has confirmed this by approving quantitative claims of dose reduction of up to 60% for one manufacturer in their marketing efforts; it is very likely that others will follow. There is one aspect in common for all manufacturers though: IR packages so far are more or less ‘black boxes’ with hardly any information given

Figure 11. In comparison to the standard analytical approach (a), statistical iterative image reconstruction reduces pixel noise considerably without a degradation of resolution (b). Apparently, this also allows for dose reduction.
on their implementation and the algorithms involved. This is legitimate, but will hopefully change in the future. A review paper on the generic principles and approaches was published in 2012 (Beister et al. 2012) and may be helpful for interested readers. Only a very brief summary of two essential approaches is given here to illustrate the goals of IR methods.

Statistical IR methods primarily aim at noise reduction and allow either for improved diagnosis in regions of high noise or for taking exams with lowered mAs products, i.e., they allow for dose reduction. The key idea of these methods is to incorporate counting statistics of the detected photons into the reconstruction process. Statistical reconstruction methods can work in the domain of sinogram space (measured raw data space), in the domain of the volumetric images or during the IR process. The first two methods working in the image domain are relatively fast compared to the third, fully iterative method, but they may also have slight negative impact on image resolution.

Model-based IR methods aim at both noise and artifact reduction and permit even greater dose reduction than statistical methods. They model the acquisition process as accurately as possible using additional information, e.g. regarding the x-ray spectrum, the size of the focal spot or detector characteristics and pixel sizes. This allows the intermediate images created for performing the iterative updates to become more and more consistent with the acquired raw data. In addition to reducing image noise, accurate modeling of geometry may be used to improve spatial resolution and modeling of the x-ray spectra allows reducing artifacts, e.g. beam hardening or metal artifacts. Since the physical acquisition process is not a trivial one, its modeling is challenging and is limited in practice by significantly increased computational demands, even with the currently increased computational power. In consequence, alternative approaches with lower computational demands such as multi-dimensional adaptive filtering as a single pre-processing step in the raw-data domain (Kachelrieß et al. 2001) or the use of PICCS (Chen et al. 2008) are of high interest and are under continued investigation.

More efficient x-ray detectors. Detector systems used in CT have often been said to provide ‘100% absorption efficiency’. This assumption is not realistic. Even for a ray hitting the center of a single detector element, the crystal’s absorption efficiency for 120 kV after penetration of 20 cm of water will typically be around 90% only; for higher kV values and for larger objects it may drop further down to about 75% even for modern detectors (Kalender 2011). Additionally, geometric efficiency has to be considered because the single elements in the detector system are separated by gaps to avoid scintillation light cross-talk between elements. This ‘dead space’ can reduce overall detection efficiency by 20%–30% and will increase further when using smaller detector pixels. Other noise sources such as electronic noise have to be considered, as discussed below. All this means that, although modern CT detector systems have been improved largely over the years, there still are further possibilities for improvement.

Some design improvements have just recently been provided. Consequent miniaturization of the electronics in the data acquisition system and the resulting omission of connecting cables, which tend to pick up noise, have led to a significant reduction of electronic noise contributions. This is of increasing importance since CT is carried out at lower and lower doses, meaning that signals are lower and that reduction of electronic noise is of increasing importance.

Great promise for detector technology is offered in two respects by so-called directly converting sensor materials. Materials such as cadmium telluride convert the energy of absorbed x-ray photons directly into electric charge thereby avoiding the signal spread encountered with the light emitted by the commonly used scintillators. In consequence, no optical separation of detector elements is required, and geometric efficiency approaches the desired 100%. The charges generated can be collected within nanoseconds which makes it possible to count single photons and to determine the energy of each photon. Instead of
analogue integration of intensities, digital counts of photons result with little or no influence of electronic noise. Single-photon counting energy-discriminating detectors have been discussed as the ultimate solution for CT over two decades already. There are some promising laboratory results by now (Kappler et al 2012) but no product solutions. First clinical applications may become possible with dedicated CT scanners such as scanners for the breast; they clearly aim at sub-mSv effective dose and at organ dose values below 5 mSv (Kalender et al 2012b).

4.3. Summary

Effective dose values today are around 1–10 mSv effective dose. Dose optimization efforts have been very successful lately due to innovative technological solutions and physics-based improvements. The downward trend in average effective dose per CT examination appears to continue; sub-mSv CT has become a realistic option, e.g. for cardiac and for pediatric CT (Kalender 2011, Cademartiri et al 2013, Schubaeck et al 2013) (figure 12).

5. Summary and outlook

CT is a technically mature imaging modality offering impressive diagnostic capabilities as is generally acknowledged. Along with the increasing clinical use of CT, special and continued scrutiny with respect to dose issues is indicated. It appears that the situation is well under control today both for scanner and for patient dosimetry. These topics have to be separated to avoid confusion and possible misconceptions as happened in the past: CTDI, the scanner exposure given in mGy mAs$^{-1}$, cannot be interpreted as patient dose.

Scanner dosimetry is well regulated and enforced for clinical CT scanners. CTDI measurements are established for the tasks of acceptance and constancy testing and for comparing scanners and scan protocols. New or longer phantoms are not required for routine tasks thus keeping CTDI measurements practical and easy to perform. There are other issues with CTDI which still deserve attention such as tests and quality control for TCM and AEC systems and dose characterization of new scanner types such as flat-detector CT, C-arm CT, or small cone-beam CT scanners. The numbers of these types of dedicated scanners are significantly increasing worldwide. Dosimetry and image quality control issues still have to be clarified (Kyriakou et al 2011); they can be handled in analogous fashion as for clinical CT.
Patient dosimetry is generally based on MC calculations calibrated by CTDI measured in air and related to anthropomorphic voxel phantoms or even the patient data itself. Measurements of CTDI in phantoms are not necessary for that purpose. Respective procedures have been in use for more than two decades and have led to the generally used practice of estimating effective dose via the DLP. It is a practical measure and serves the general purpose of dose surveillance well although it cannot be used for estimation of individual patient dose.

Estimates of organ doses pose a greater challenge and should be specific for the patient, the anatomic scan range, the scanner and the scan protocol under consideration. Both professionals and the general public expect that accurate data can be made available; organ dose estimates, not effective dose $E$, are the primary goal in many cases. Such values can be provided by the POSDE concept outlined above. The fast MC programs available today (Chen et al 2012) using hardware acceleration allow for POSDE in acceptably short times of 2–10 min. Nevertheless, such specific evaluations appear indicated only for special cases or scientific studies. High accuracy is possible but can only be guaranteed if manufacturer cooperation is ensured.

Dose consciousness is very high in CT practice today. Respective efforts have meanwhile led to substantial dose reduction, and sub-mSv CT has become a realistic and routine option for a number of applications. The trend is expected to continue, and the ALARA principle is certainly respected. But there is a necessary caveat: Do not go too low in dose! Optimization demands that adequate diagnostic quality be reached; i.e., the right dose level has to be applied so that diagnostic accuracy is not compromised. The diagnostic benefits for the patient are acknowledged in general; for example, CT Imaging was cited as one of the most significant medical innovations in the previous decades in a US survey (Fuchs and Sox 2001). Benefits appear to be increasing further in view of the many technological advances and improved applications.

The risk of x-ray examinations in the low-dose range is not well known at all; no hard data are available for the dose range below 100 mSv (Feinendegen et al 1999, Tubiana et al 2009, Hendee and O’Connor 2012, Neumaier et al 2012). Nevertheless cancer deaths are often calculated based on the linear no-threshold (LNT) hypothesis although relevant scientific societies clearly speak against it. The situation was analyzed and summarized recently in a concise fashion by Hendee (2013). If we use the LNT hypothesis nevertheless, the fatal cancer risk of 5% per Sv of effective dose has to be assumed (ICRP 2007). This means that one person out of 50 000 exposed by 1 mSv is affected. Risks of that magnitude are generally termed ‘very low’; exposures between 2 and 20 mSv are considered ‘low’ risks (Calman 1996, ARPANSA 2005, Hendrick et al 2012). For comparison, life time risks of death by bicycle accidents, motor vehicle accidents and cancer from natural causes of 1:10,000, 1:100 and 1:5, respectively, have been determined; these data are not hypothetical but based on solid statistics (Hendrick et al 2012).

There is no situation in life where a ‘zero risk’ can be guaranteed; low or hypothetical risks are generally acceptable if there is a substantial benefit. The benefit-to-risk ratio is the essential quantity and should be ‘as high as reasonably achievable (AHARA)’. AHARA was and is the goal for further work on dose issues in CT. Continued and rigorous dose surveillance is a necessity. With that ensured, CT can be a low- or even a very low-dose modality.

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