Radiation Exposure in Computed Tomography

Fundamentals, Influencing Parameters, Dose Assessment, Optimisation, Scanner Data, Terminology

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Collective Effective Dose (in Germany)

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Preface

Only a few months after its release, Strahlenexposition in der Computertomographie, the first edition of this book published in the German language with the support of the Medical Engineering Division of the German Electrical and Electronics Manufacturers’ Association, ZVEI, has rapidly become a well-recognized and useful work in its field. A major key to this success was the support and commitment of all manufacturers of computed tomography (CT) equipment in Germany, who supplied their customers with this tool so as to overcome a serious educational problem in this specific, but important area of diagnostic radiology.

Although the awareness of radiation exposure associated with CT is not uniform all over Europe, the problem is everywhere similar: a single imaging modality that represents a small component in the spectrum of radiological procedures, but contributes greatly to the population dose arising from medical exposures. This will become even more evident in the near future as diagnostic reference levels are currently established all over Europe as a consequence of the European directive on the radiation protection of patients. This is a new approach which allows for benchmarking and which requires firm actions to optimise CT exposure practices at many installations.

The rapid development of magnetic resonance imaging (MRI), which is not tainted with the adverse effects of ionizing radiation, does indeed offer powerful alternatives to CT in several areas. Nevertheless, CT is a well-established and cost-effective diagnostic modality which has attracted additional interest following the recent introduction of multi-slice CT. Multi-slice CT, however, has not only brought new possibilities for application, but also new dose-related operational factors that require additional attention. Therefore the reduction of radiation exposure to the level which is absolutely necessary from a diagnostic point of view is of key importance to preserve the reputation of CT and to maintain its position both today and in the future.

When the first German edition was published, the editor of Strahlenexposition in der Computertomographie was encouraged from various sides to provide an English translation of this book. As in the previous case, the broad support of CT manufacturers has enabled the realization of these requests and the presentation of a revised English edition in cooperation with the European Coordination Committee of the Radiological and Electromedical Industries (COCIR). Recent developments and findings, such as the results from a nationwide survey on CT exposure practices in Germany that were not available when the previous edition was published, can now be incorporated.

This study, which was carried out as a collaboration between the German Roentgen Society (DRG) and ZVEI, is a good example of how successful the German Concerted Action Dose Reduction in CT has already been since its foundation in 1998. The financial and personal support given to this and other projects by the CT manufacturers, underlines their determination to contribute within their power to provide solutions for this challenging task.

COCIR is happy to present - just in time - this important contribution towards dose control in this essential area of diagnostic radiology. On behalf of the European electromedical industry, COCIR is greatly indebted to the editor and the co-authors of this book for having treated this complex subject in a comprehensive, but compact and illustrative manner. We hope that this work will find similar widespread use and receive similar recognition as its German predecessor.

Frankfurt am Main, autumn 2000

Hans-Peter Bursig

COCIR Secretary General
The importance of computed tomography as an imaging modality has grown rapidly in the past 25 years since its introduction. At present, CT is recognized as having the highest diagnostic value among radiological procedures. However, for a long time there was a lack of awareness of the radiation exposure associated with CT. Its full extent in Germany became evident only after a survey which was conducted by the Federal Bureau on Radiation Protection in 1994: while CT comprises only 4% of all radiological procedures, it gives rise to approximately 1/3 of the population dose from medical exposures. In view of these facts and an ongoing tendency for increasing frequencies of CT examinations, CT represents by far the greatest challenge in the field of radiation protection in medicine.

The manufacturers of CT equipment, organized in the Medical Engineering Division of the German Electrical and Electronics Manufacturers’ Association, ZVEI, share the concern of this development and feel themselves called upon to contribute within their power to dose reductions in the field of CT. The reasons that have led to the relatively high dose levels which are commonly used today, however, are quite complex. Solutions based on isolated, individual measures are therefore not very promising. Significant reduction of radiation exposure in CT will only be achieved by joint efforts of all the parties involved.

In spring 1998, ZVEI therefore started an initiative by inviting all parties to a discussion forum held at Frankfurt. At this meeting, the Concerted Action Dose Reduction in CT was founded which has since then been conducted by the German Roentgen Society. Subsequent discussions have shown that there is a significant deficit in the field of training, where a lack of appropriate courses and literature is a matter of fact.

This book is seen as a first brick to reduce this deficit. In collaboration with well-known radiologists and medical physicists from a range of universities, Dr. H. D. Nagel, technical delegate of the X-ray & CT working group of the Medical Engineering Division of ZVEI, has produced a compendium which offers a comprehensive overview of all relevant questions related to radiation exposure in CT. On behalf of the CT manufacturers organized in the Medical Engineering Division, ZVEI expresses thanks to the team of authors for their qualified and practically oriented contributions to this book and hopes that the efforts undertaken by the authors will help very soon to improve the situation.

Frankfurt am Main, autumn 1999

Ellen-Urs Meyer-Schülke
Secretary of
Medical Engineering Division of
ZVEI e.V
# Table of Contents

Chapter 1: A Few Remarks on Radiation Exposure in CT ................................................................. 1

Chapter 2: Fundamentals of CT Dosimetry ......................................................................................... 5

- Differences between CT and Conventional Projection Radiography ........................................ 5
- Dose Quantities Appropriate for CT .............................................................................................. 6
  - Computed Tomography Dose Index (CTDI) .............................................................................. 6
  - Average Dose and Weighted CTDI ............................................................................................ 9
  - Effective CTDI .......................................................................................................................... 9
  - Dose Free-in-Air on the Axis of Rotation .................................................................................. 10
  - Organ Dose ............................................................................................................................... 10
  - Dose-Length Product (DLP) ...................................................................................................... 11
  - Effective Dose .......................................................................................................................... 13
  - Summary of CT Dose Quantities .............................................................................................. 13

Chapter 3: Dose Values from CT Examinations ................................................................................. 15

- Methods for the Determination of Effective Dose ...................................................................... 15
  - Order of Magnitude Estimation .............................................................................................. 15
  - Calculations Based on Tabulated Conversion Coefficients .................................................... 16
  - Effective Dose for Paediatric Examinations .......................................................................... 17
  - Estimation of Effective Dose from DLP ................................................................................. 17
  - Scanner-specific Corrections ................................................................................................. 18
  - Corrections for the Influence of Tube Potential ...................................................................... 19
  - Dose from Scan Projection Radiography ............................................................................... 19
  - Summary of Principle for Dose Calculation ........................................................................... 19
  - Dose Calculation Software .................................................................................................... 19

- Results from Dose Surveys ........................................................................................................ 21
  - Why is Radiation Exposure Increased in CT? .................................................................... 23

Chapter 4: Factors Influencing Patient Dose in CT ........................................................................ 25

- Equipment-related Factors .......................................................................................................... 25
  - Waveform of the Generator ................................................................................................. 25
  - Range of Tube Current Settings ............................................................................................ 25
  - Beam Filtration ....................................................................................................................... 26
  - Beam Shaper .......................................................................................................................... 26
  - Focus-Axis Distance ............................................................................................................. 26
  - Slice Collimation .................................................................................................................... 27
  - Detector Array ....................................................................................................................... 27
  - Scan Geometry (Generation) .............................................................................................. 28
  - Partial Fan Beam Scanners ................................................................................................. 29
  - Scan Angle ............................................................................................................................. 29

- Application-related Factors ......................................................................................................... 30
  - Relationship between Dose and Image Quality .................................................................... 30
  - Tube Current-Time Product (Q) ......................................................................................... 31
Table of Contents

<table>
<thead>
<tr>
<th>Topic</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Exposure Time (t)</td>
<td>31</td>
</tr>
<tr>
<td>Tube Potential (U)</td>
<td>32</td>
</tr>
<tr>
<td>Object Diameter (d)</td>
<td>32</td>
</tr>
<tr>
<td>Slice Thickness (h)</td>
<td>33</td>
</tr>
<tr>
<td>Pitch Factor (p)</td>
<td>34</td>
</tr>
<tr>
<td>Number of Slices (n)</td>
<td>34</td>
</tr>
<tr>
<td>Reconstruction Filter (FK)</td>
<td>34</td>
</tr>
<tr>
<td>Window Width (W)</td>
<td>35</td>
</tr>
<tr>
<td>Matrix Size and Field-of-View (FOV)</td>
<td>36</td>
</tr>
<tr>
<td>Summary of Application-related Factors</td>
<td>37</td>
</tr>
<tr>
<td>Aspects of Dose for Special Technical Features</td>
<td>37</td>
</tr>
<tr>
<td>Spiral CT</td>
<td>37</td>
</tr>
<tr>
<td>Multi-Slice Systems</td>
<td>38</td>
</tr>
<tr>
<td>Tube Current Modulation</td>
<td>42</td>
</tr>
<tr>
<td>CT Fluoroscopy</td>
<td>42</td>
</tr>
<tr>
<td>Scientific Studies on Low-Dose CT</td>
<td>45</td>
</tr>
<tr>
<td>Low-Dose CT of the Chest</td>
<td>45</td>
</tr>
<tr>
<td>LD-HRCT of the Chest</td>
<td>47</td>
</tr>
<tr>
<td>Low-Dose CT in Paediatric Examinations of the Pelvis</td>
<td>48</td>
</tr>
<tr>
<td>Low-Dose CT of the Skeleton</td>
<td>48</td>
</tr>
<tr>
<td>Applications and Limitations of Low-Dose CT</td>
<td>49</td>
</tr>
<tr>
<td>Potential Means for Dose Reduction</td>
<td>51</td>
</tr>
<tr>
<td>Chapter 6: Dose Measurements, Reference Values, Examples</td>
<td>55</td>
</tr>
<tr>
<td>Dosemeters and Dosimetry Phantoms</td>
<td>55</td>
</tr>
<tr>
<td>Measurement Procedure</td>
<td>56</td>
</tr>
<tr>
<td>Measurement of CTDI Free-in-Air</td>
<td>56</td>
</tr>
<tr>
<td>In-Phantom Measurement of CTDI</td>
<td>57</td>
</tr>
<tr>
<td>Dose Limits and Reference Values</td>
<td>58</td>
</tr>
<tr>
<td>BAK Guideline on CT</td>
<td>58</td>
</tr>
<tr>
<td>Guideline for Acceptance Testing (Regelwerk 13)</td>
<td>58</td>
</tr>
<tr>
<td>European Quality Guidelines for CT</td>
<td>59</td>
</tr>
<tr>
<td>National Reference Dose Values for CT</td>
<td>59</td>
</tr>
<tr>
<td>Examples for Dose Assessments</td>
<td>60</td>
</tr>
<tr>
<td>Example 1: Compliance with Dose Limits Given in Terms of CTDI Free-in-Air</td>
<td>60</td>
</tr>
<tr>
<td>Example 2: Compliance with EC Recommendations</td>
<td>60</td>
</tr>
<tr>
<td>Example 3: Effective Dose for a Standard CT Examination of the Head</td>
<td>61</td>
</tr>
<tr>
<td>Example 4: Effective Dose for a Spiral CT Examination of the Chest</td>
<td>61</td>
</tr>
<tr>
<td>Example 5: Effective Dose for a Spiral CT Examination of the Chest of a Baby</td>
<td>62</td>
</tr>
<tr>
<td>Example 6: Effective Dose for a CT Examination Performed Using a Compact Scanner</td>
<td>62</td>
</tr>
<tr>
<td>Example 7: Calculation of Effective Dose from the Dose Reading Displayed at the Console</td>
<td>63</td>
</tr>
<tr>
<td>Example 8: Effective Dose for a CT Examination Performed Using a Multi-Slice System</td>
<td>63</td>
</tr>
</tbody>
</table>
Chapter 1: A Few Remarks on Radiation Exposure in CT

Th. Schmidt

The world-wide fascination in 1972 about the possibility of producing non-superimposed, cross-sectional images was probably only exceeded by the earlier enthusiasm when X-rays were discovered in 1895. The ‘discovery’ of CT, however, didn’t occur accidentally. Rather, it was the realisation of an ancient dream.

Surprisingly, the breakthrough was not achieved by one of the major X-ray imaging companies but by an outsider in this field, the British recording company EMI. The principle presented by EMI was not unknown to the manufacturers of X-ray equipment. However, the major companies had all been mislead by their desire to exceed the spatial resolution achievable with all existing X-ray imaging equipment. This possibility was not, and absolutely is not, the case for CT, especially for the first prototypes. The major X-ray imaging companies simply ignored the fact that any improvement in contrast resolution can provide important additional information - even if spatial resolution is poor.

At the beginning of CT history, scanners were exclusively used for head examinations. The first CT scanner already allowed the differentiation of haematomas, infarcts and tumours from normal brain tissue without the administration of contrast agents. The enthusiasm about these new opportunities was world-wide and thus explains the rapid development of CT (Dümmling84). In 1974 the first body scanners became available. The classical X-ray manufacturers were now in the race, too. The subsequent rapid development is illustrated in fig. 1.1 and table 1.1. CT very soon became an established and indispensable imaging modality.

**Tab. 1.1**
Number of CT units installed and CT scanner density in various European countries in 1998 as estimated by COCIR. CT scanner density varies markedly depending on the health care system (reference: COCIR99).

<table>
<thead>
<tr>
<th>Country</th>
<th>CT units</th>
<th>Population (Mio.)</th>
<th>CT density (per Mio.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Belgium</td>
<td>258</td>
<td>10.2</td>
<td>2.5</td>
</tr>
<tr>
<td>Finland</td>
<td>50</td>
<td>5.0</td>
<td>1.0</td>
</tr>
<tr>
<td>France</td>
<td>611</td>
<td>58.4</td>
<td>1.1</td>
</tr>
<tr>
<td>Germany</td>
<td>1863</td>
<td>81.7</td>
<td>2.3</td>
</tr>
<tr>
<td>Italy</td>
<td>1182</td>
<td>57.3</td>
<td>2.1</td>
</tr>
<tr>
<td>The Netherlands</td>
<td>164</td>
<td>15.0</td>
<td>1.1</td>
</tr>
<tr>
<td>Spain</td>
<td>546</td>
<td>39.3</td>
<td>1.4</td>
</tr>
<tr>
<td>Sweden</td>
<td>122</td>
<td>8.8</td>
<td>1.4</td>
</tr>
<tr>
<td>Switzerland</td>
<td>180</td>
<td>7.1</td>
<td>2.5</td>
</tr>
<tr>
<td>United Kingdom</td>
<td>387</td>
<td>57.0</td>
<td>7.0</td>
</tr>
<tr>
<td><strong>TOTAL</strong></td>
<td><strong>5183</strong></td>
<td><strong>339.8</strong></td>
<td><strong>15</strong></td>
</tr>
</tbody>
</table>

The perfection of the CT technique was accompanied by a considerable reduction in scan time. At the beginning of the CT era, scan and reconstruction times amounted to approximately 8 minutes for a single slice. Nowadays, less than 1 second is needed instead. Retrospectively, the number of slices per examination turns out to depend on the state-of-the-art in scanning technology.

**Fig 1.1**
Growth in the number of CT scanners installed in hospitals and private practices in Germany (references: Schmidt93, ZVEI99).
Chapter 1: A Few Remarks on Radiation Exposure in CT

and the time needed to carry out one scan. Fig. 1.2 illustrates this trend.

The number of CT examinations has of course also increased as a consequence of the growing number of installed scanners. In Germany, for example, the annual number of CT examinations per scanner varies between 3500 and 7000, with the CT scanners installed in hospitals being used more frequently than those in private practices. On the basis of estimates for the numbers of installed CT scanners, examinations per scanner and slices per examination, it is possible to project the total number of CT slices. This amounts to some 100 million per year, which is in the same order of magnitude as the sum of all the images produced using conventional projection radiography. Fig. 1.3, (which is based on the number of examinations, not on the number of slices and images, respectively!) illustrates the role that CT plays in the spectrum of diagnostic X-ray applications.

When CT was first introduced, the question of the radiation exposure associated with this new technique did arise, of course. In the beginning, however, the primary concern in CT was to define strategies for examinations and to develop standards for scanning protocols, thereby attributing a lower priority to dose aspects. Only at the end of the 1980s did a larger number of CT users become aware of the considerable exposure associated with CT. In the UK, a nation-wide survey on CT practice was conducted by the National Radiological Protection Board NRPB (Shrimpton91). In Germany, CT guidelines were published in 1992 by the German Federal Chamber of Physicians (BÄK92) that were intended to draw attention to this problem and to set up dose recommendations which should not normally be exceeded.

An estimation of the contribution from CT to the radiation exposure of patients is shown in fig. 1.4, based on data from a survey on the frequency of radiological examinations in Germany which was conducted by the Federal Bureau on Radiation Protection (BfS) during the period 1990 - 92 (Bernhardt95). The result for CT is surprising in that such a relatively small fraction of only 4% of all examinations (fig 1.3) can give rise to a percentage contribution to the collective effective dose which is one order of magnitude higher. These figures were confirmed in a recent, nation-wide survey conducted in Germany. CT therefore constitutes by far the largest contribution to the radiation exposure of the population from diagnostic medical sources. Undoubtedly, similar situations are known or can be assumed to exist in all other industrialized countries.

**Fig. 1.2**

Developments in the number of slices per examination with the slice thickness (in mm) given in brackets (reference: Schmidt93).

**Fig. 1.3**

Percentage frequencies of X-ray examinations in the former FRG during the period 1990 - 92 (reference: Bernhardt95).
Fig. 1.4
Percentage contributions from various X-ray examinations to the collective effective dose in the former FRG during the period 1990 - 92 (reference: Bernhardt95).

Even if there might be some uncertainties in the data in fig. 1.4, the general trend is clear. Savings in collective dose over the past few years achieved, for example, by introducing film-screen combinations of higher speed index, will have been compensated by the increasing contribution resulting from CT examinations. The fact that such a small fraction of X-ray examinations can contribute so greatly to population dose must set everybody thinking.

New CT techniques such as spiral CT, electron beam CT or multi-slice CT will not behave much differently in terms of patient dose as long as similar examination protocols are used (scan length, slice thickness etc.). The increased speed and the ease of use for scanning long distances, however, rather increase the radiation exposure from a particular examination. This automatically raises the question of the influence of the responsible radiologist on the exposure associated with a certain examination.

A potential for dose reduction can primarily be assumed for areas such as

- automatic adjustment of scan parameters to patient size,
- definition of, and commitment to, a diagnostically adequate image quality,
- development of a catalogue of indications for CT, and
- improved training for the users of CT.

In order to solve these problems, close co-operation is mandatory between the developers of CT equipment, the training staff of CT manufacturers, radiologists, scientific societies and legal authorities. The publication of reference dose values for a number of standard CT examinations should automatically lead to the optimization of scan protocols. The ‘education’ of CT users by the training staff of CT manufacturers and dedicated training courses, however, will most probably result in a more significant reduction of doses in CT.

The aim of this book is to fill a gap in knowledge by both introducing all parties which are involved in the CT business into the fundamentals of CT-related dosimetry and also demonstrating the mutual dependencies between dose and image quality. In order to achieve this goal, the essential differences between CT and conventional projection radiography are illustrated and dose quantities appropriate for CT are presented in chapter 2. Chapter 3 describes various methods for dose assessment from the knowledge of scan parameters and tries to answer the question of why CT is more dose-intense than conventional procedures. In chapter 4, the impact of technical and operational parameters on radiation exposure is discussed; furthermore, special techniques (e.g. spiral CT, CT fluoroscopy etc.) are considered with respect to aspects of dose.

Examples of how low-dose CT techniques can be applied to selected types of examinations in daily routine are given in chapter 5. Chapter 6 presents methods and equipment for dose measurements, reference dose values and dose limits, and a number of arithmetical problems on how to make dose assessments. The themes of this book are completed by an appendix with numerical data relevant to specifications of dose for most CT scanners and a comprehensive glossary, in which all dose-related terms are briefly described.

From the perspective of the authors of this book, the hope remains that this work will find widespread distribution and acceptance, thus providing a contribution to the reduction of radiation exposure in CT without sacrificing diagnostic confidence.
Chapter 2: Fundamentals of CT Dosimetry

H.D. Nagel

CT and dose - far too often a book of mystery for many of those who have to deal with this imaging modality. When asking somebody for the radiation exposure from a given CT examination, the casual answer is: ‘so and so many mAs’. Even in scientific publications the applied tube current-time product is used as a synonym for radiation exposure (e.g. Hoe98).

This point of view is not completely wrong as there is a linear relationship between the applied tube current-time product and radiation dose. However, it is often not recognized that this relationship differs depending on the type of scanner. Dose comparisons in terms of mAs statements are therefore not appropriate in the field of CT and are far from allowing a reasonable indication of the radiation exposure relative to that from conventional X-ray projection techniques.

What are the reasons for these difficulties found in CT dosimetry? There is at least a vague impression that the quantities and measuring procedures which are well established in conventional radiology cannot be easily translated, because of the obvious differences between the transverse scanning technique used in CT and the projection technique in conventional radiography. But which terms are appropriate for the description and assessment of radiation exposure in CT instead? And shouldn’t there be some common denominators which would allow the comparison of radiation exposures from CT procedures with those from conventional X-ray examinations?

What are the differences between conventional projection radiography and CT? How are dose quantities defined that are suitable for CT? What is the meaning of these quantities and how well do they characterize radiation exposure in CT? Answers will be given to these and a number of other related questions in this chapter.

Differences between CT and Conventional Projection Radiography

The dose distribution inside the patient in CT is completely different from that for a conventional radiogram. In conventional projection radiography, the dose decreases continuously from the entrance of the X-ray beam to its exit, with a ratio of between 100 and 1000 to 1. In the case of CT, with the rotational geometry that is now common to all scanners, the dose is almost equally distributed in the scanning plane. This is primarily a consequence of the scanning procedure, in which the patient is equally irradiated from all directions during a complete rotation of the X-ray tube. Therefore the dose is concentrated at the centre of rotation despite the attenuation of the surrounding tissue. This situation resembles the rotational irradiation that is used in radiation therapy to provide a favourable ratio between dose at the desired depth and the inevitable dose at the surface of the body. A dose comparison of CT with conventional projection radiography in terms of skin dose therefore doesn’t make any sense.

Another fundamental difference arises from the circumstances in which CT examinations are performed by imaging transverse slices of the body that are only a few millimetres thick. As the dose profile shown in fig. 2.1
Chapter 2: Fundamentals of CT Dosimetry

demonstrates, radiation energy is deposited not only in the slice itself, but also in its neighbourhood. Apart from the limited efficiency of beam collimation (penumbra) and the divergence of the radiation beam, this phenomenon is mainly caused by scattered radiation that is produced in the slice. The nominal slice thickness, \( h \), is approximately equivalent to the full width at half maximum of the dose profile, unless the beam is further collimated close to the detector. The tails of the dose profile therefore contribute significantly to dose even outside the irradiated slice.

Finally, the situation in CT is further complicated by the circumstances in which - unlike in conventional projection radiography - the volume to be imaged is not irradiated simultaneously. Instead, a CT examination consists of a series of single slices which, in conventional serial scanning technique, are acquired sequentially in order to cover the entire volume stepwise (special aspects of spiral scanning techniques will be treated separately in chapter 4). This often leads to confusion about what the

Dose Quantities Appropriate for CT

In the following sections dose quantities which are appropriate for CT are described, analyzed and - wherever possible - put in relation to each other. In this context it has turned out as very useful to distinguish between local and integral dose quantities. Local dose quantities are indicators of the intensity of the irradiation inside the limits of the irradiated body region. Computed Tomography Dose Index (CTDI), dose free-in-air on the axis of rotation and - with some restrictions - organ dose are members of this group, for which the general term ‘dose’ is used in this book. In contrast, integral dose quantities, such as dose-length product and effective dose, are descriptors of the total amount of radiation absorbed by taking into account also the extent of the body region being irradiated. These integral dose quantities will be addressed in this book by using the general term ‘radiation exposure’.

Computed Tomography Dose Index (CTDI)

The Computed Tomography Dose Index (CTDI) has for many years commonly been used as the most-specific dose quantity for CT. Whenever - as usually occurs in practice - several adjacent slices are scanned instead of a single slice, the dose for a particular slice is increased due to the contributions from slices in its neighbourhood. In fig. 2.3 on the left, the total dose profile is illustrated for an examination consisting of 15 slices (nominal slice thickness 10 mm, table feed equal to slice thickness (pitch = 1), dose profile as shown in fig. 2.2).

---

**Fig. 2.2**
Typical dose profile for a single slice with a nominal slice thickness \( h = 10 \text{ mm} \).

dose from a complete series of e.g. 15 slices might be compared with the dose from a single slice.

**Fig. 2.3**
Total dose profiles for two scan series consisting of several slices, each 10 mm thick (MSAD = Multiple Scan Average Dose).
Left: examination with 15 slices and 10 mm table feed (pitch = 1); right: examination with 21 slices and 7 mm table feed (pitch = 0.7).
By superimposition of all of these single dose profiles, the dose in the central portion of the total dose profile increases in this example to a level that is 1.5 times the peak value for a single slice. In the literature, this increased value is called the ‘Multiple Slice Average Dose (MSAD)’ (Shope81).

If the examination is performed with overlapping slices, i.e. by using a table feed smaller than the slice thickness, the increase in dose becomes even larger. The packing factor is used as an indicator of the degree of overlap, which is roughly given by the ratio of slice thickness and table feed. If overlapping slices are used, the packing factor is always > 1. Instead of the packing factor, the term pitch is often used. Pitch is the inverse of the packing factor and is defined as

\[
\text{Pitch } p = \frac{\text{Table feed TF}}{\text{Slice thickness } h}
\]

The example given on the right in fig. 2.3 shows the situation which results from using 10 mm thick slices and a table feed of only 7 mm (p = 0.7). The dose level in the central portion of the irradiated region is increased to 2.2 times the peak value for a single slice with the dose profile as shown in fig. 2.2.

But what else is CTDI? Fig. 2.4 illustrates the meaning of this term: CTDI is the equivalent of the dose value inside the irradiated slice that would result if the absorbed radiation dose profile were entirely concentrated to a rectangular profile of width equal to the nominal slice thickness. Accordingly, all dose contributions from outside the nominal slice width, i.e. the areas under the tails of the dose profile, are added to the area inside the slice.

The corresponding mathematical definition of CTDI therefore describes the summation of all dose contributions along a line which is parallel to the axis of rotation for the scanner (= z-axis):

\[
CTDI = \frac{1}{h} \cdot \int_{-\frac{h}{2}}^{\frac{h}{2}} D(z) \cdot dz
\]

where \( D(z) \) is the value of the dose at a given location, \( z \), and \( h \) is the nominal slice thickness. CTDI is therefore equal to the area of the dose profile (the ‘dose-length product’) divided by the nominal slice thickness.

Since the increased level in dose for the MSAD results from the tails of the single dose profiles for a scan series, it is obvious that MSAD and CTDI are exactly equal if the following conditions are fulfilled: table feed equal to slice thickness (i.e. pitch \( p = 1 \)) and a sufficient number of slices in order to reach the saturation level of MSAD. The latter is the case after 10 to 12 slices (Shope81).

In general (i.e. if \( p \) is not equal to 1), the relationship between CTDI and MSAD is given by

\[
MSAD = \frac{1}{p} \cdot CTDI
\]

The practical implication of equation (2.3) is that - in order to obtain the average dose for a scan series - it is not necessary to carry out all the scans. Instead, it is sufficient to obtain the CTDI from a single scan by acquiring the entire dose profile according to equation (2.2).

The definition most commonly used for statements of CTDI values restricts the summation of dose contributions to a length of ±7 slices on either side of the irradiated slice, i.e. to a total length equal to 14 times the nominal slice thickness:

\[
CTDI_{FDA} = \frac{1}{h} \cdot \int_{-\frac{7}{2}h}^{\frac{7}{2}h} D(z) \cdot dz
\]

The suffix ‘FDA’ tells us that this kind of CTDI measurement refers to the wide-spread definition of CTDI according to requirements for CT scanners in the USA (DHHS84). The dose quantity \( D \) according to FDA is absorbed dose to Perspex (also known as Lucite, Plexiglas or PMMA). In order to fulfil US requirements, manufacturers of CT scanners are obliged to report CTDI values according to equation (2.4) for all modes of operation.
With decreasing slice thickness, the length over which the dose contributions are summed becomes shorter and shorter in the case of CTDI_{FDA}. However, the length of the tails on the dose profile is certainly not reduced to the same extent. As a consequence, the dose from slice thicknesses smaller than 7 mm is progressively measured less completely for decreasing settings of slice thickness. To avoid this systematic underestimation of dose, the definition of CTDI has been modified in such a way (Leitz95, IEC99) that a fixed integration length of 100 mm is used, independent of slice thickness:

$$CTDI_{100} = \frac{1}{h} \cdot \int_{-50}^{50} K_{air}(z) \cdot dz$$  \hspace{1cm} (2.5)$$

This kind of CTDI is sometimes referred as the ‘Practical CTDI (PCTDI)’ (Leitz95, Scheck98). Instead of absorbed dose to Perspex, air kerma $K_{air}$ is the dose quantity used for CTDI_{100}. It is therefore no longer necessary to convert from air kerma to absorbed dose to Perspex. Thus, a potential source of error of approximately 10% in CTDI is removed, since this conversion may not always have been made correctly.

Most of the older CTDI statements in the literature and specification sheets were made in terms of CTDI_{FDA}. If it is not explicitly stated which definition of CTDI has been used, additional information should be sought. As a general indication, CTDI values become smaller with decreasing slice thickness, if based on CTDI_{FDA}. Approximate correction factors to convert from CTDI_{FDA} to CTDI_{100} are given in table 2.1. These factors also include the conversion from absorbed dose to Perspex into air kerma.

It can be concluded from table 2.1 that for 10 mm thick slices not much is changed in practice. The effect of changing integration length (which is slightly smaller than necessary in this case) is almost completely compensated by the correction for using a different dose quantity. In the case of thinner slices, however, the increasing value of the correction factors demonstrates the extent to which CTDI_{FDA} has underestimated the real situation and how useful it was to introduce a more relevant quantity such as CTDI_{100}.

In the standard measuring procedure for CTDI, which utilizes two cylindrical Perspex phantoms of different diameter, dose values are obtained at the centre and near the periphery of the phantom. The larger phantom, being 32 cm in diameter, represents the absorption that is typical for the trunk region of adults. The smaller phantom (16 cm in diameter) represents the patient in head examinations. The smaller phantom is also used for dose assessment in pediatric examinations. Further details related to the measuring procedure can be found in chapter 6.

CTDI statements in scanner specification sheets are given for the head phantom as well as for the body phantom, and often apply to a current-time product of 100 mAs or 1 mAs. In this case it must be recognized that a quantity named ‘normalized CTDI’ is used, which is labelled $\alpha_{CTDI_{FDA}}$ or $\alpha_{CTDI_{100}}$ in order to avoid confusion. The normalized CTDI is obtained by dividing the CTDI value by the mAs product $Q$ that was used to measure CTDI:

$$\alpha_{CTDI_{XYZ}} = \frac{CTDI_{XYZ}}{Q}$$  \hspace{1cm} (2.6)$$

It is worthwhile (and indeed necessary) to note that the normalized CTDI is a characteristic quantity for a scanner (dose rate coefficient) which simply represents the capacity of a scanner in terms of output and which conveys absolutely nothing about patient dose. Very often it is assumed that scanners with a high value of $\alpha_{CTDI}$ are more ‘dangerous’ than other models with lower $\alpha_{CTDI}$ values. This is not necessarily the case. Reference to patient dose can not be made unless the normalized CTDI has been multiplied by the tube current-time product $Q$ that is required in order to produce images of diagnostic quality with the type of scanner under consideration:

$$CTDI_{XYZ} = \alpha_{CTDI_{XYZ}} \cdot Q$$  \hspace{1cm} (2.7)$$

Only after having carried out this step is it possible to decide if a particular scanner delivers more or less dose than another model for a specified type of examination.

Usually pairs of CTDI values are given for each phantom size, one value related to the centre of the phantom

<table>
<thead>
<tr>
<th>Slice thickness (mm)</th>
<th>16 cm head phantom</th>
<th>32 cm body phantom</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Periphery</td>
<td>Centre</td>
</tr>
<tr>
<td>1</td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td>5</td>
<td>1.2</td>
<td>1.3</td>
</tr>
<tr>
<td>3</td>
<td>1.3</td>
<td>1.6</td>
</tr>
<tr>
<td>2</td>
<td>1.5</td>
<td>2.0</td>
</tr>
<tr>
<td>1.5</td>
<td>1.6</td>
<td>2.2</td>
</tr>
<tr>
<td>1</td>
<td>3.1</td>
<td></td>
</tr>
</tbody>
</table>

Tab. 2.1
Conversion factors to estimate CTDI_{100} from CTDI_{FDA} statements (conversion from absorbed dose to Perspex into air kerma included (reference: ImPACT00)).
(CTDI, ‘centre’) and another one related to the periphery (CTDIp, ‘edge’). An example of such normalized CTDI values is given in table 2.2. These CTDI values are typical for modern scanners with reduced geometrical dimensions (60 cm focus-to-axis distance) and beam quality (120 kV, filtration 5 to 6 mm Al-equivalent).

In case of the head phantom, the CTDI values at the centre and at the periphery are almost equal. The CTDI values for the body phantom are comparably smaller, as a result of the increased attenuation for the larger phantom diameter, whereby the CTDI value at the centre is only 50% of the corresponding value at the periphery. Nevertheless, the values for the different phantoms differ by only a factor 3 despite the considerable difference in their size. Therefore it is not necessary to supply additional CTDI values for other phantom diameters.

What does the CTDI tell us about the radiation exposure of the patient? According to its definition, CTDI only represents a measure of the intensity of irradiation at a specified location, i.e. inside the slice under consideration. Apart from the dose pile-up effect mentioned earlier, the CTDI is unable to represent the complete radiation exposure, i.e. the integral effect of the entire scan series. As the CTDI is obtained at specified points inside a phantom, however, the resulting values can be used as indicators of dose to those organs which are located near the body surface (e.g. eye lens, breast) or at the centre of the body (e.g. uterus). CTDI100 is much better suited for this purpose than the former CTDI FDA. Since only phantoms of cylindrical shape are used, the absorption conditions of differently shaped body regions, such as the shoulder and pelvis, are not very well represented.

**Average Dose and Weighted CTDI**

In the last few years, the terms ‘average dose’ and ‘weighted’ CTDI have increasingly been used (e.g. in the European quality guidelines for CT (EC99) and in the multi-centre CT study of the German Roentgen Society (Scheck98)). Weighted CTDI is a weighted mixture of the pair of CTDI100 values, with weightings of 1/3 for the central CTDI100c and 2/3 for the peripheral CTDI100p (Leitz95):

\[
CTDI_w = \frac{1}{3} \cdot CTDI_{100c} + \frac{2}{3} \cdot CTDI_{100p}
\]  

(2.8)

CTDIw must be calculated separately for both the head and body phantoms. Once again, it is important to differentiate between absolute and normalized values of CTDIw, the latter being abbreviated as nCTDIw. The only advantage of CTDIw is that it enables the use of a single number instead of two, particularly in the case of the body region where the central and peripheral values are not of the same magnitude. Apart from this, CTDIw behaves neither better nor worse as a dose descriptor for CT than the ordinary CTDI.

Future dose recommendations (‘reference dose values’) will be stated in terms of weighted CTDI. The same holds true for the dose display that will be mandatory for all new scanners in the near future (IEC99). Therefore, this quantity has a prominent role as an indicator of the local dose related to a single slice.

**Effective CTDI**

The dose display at the operator’s console which is now required for all new scanners is calibrated in terms of weighted CTDI. In those cases, however, where the pitch factor is not equal to 1, CTDIw is divided by the value of the pitch (IEC99). The display therefore shows a kind of corrected or effective weighted CTDI, which shall simply be called ‘effective CTDI’ and which is defined as

\[
CTDI_{w,\text{eff}} = \frac{1}{p} \cdot CTDI_w
\]  

(2.9)

By doing this, the impact of pitch on the entire volume of the examination (i.e. radiation exposure) is already taken into account at the level of local dose. Similarly, tube current-time products are sometimes reported in terms of an ‘effective mAs’ (e.g. Galanski98) which may differ significantly from the chosen settings of tube current and scan time.

Effective CTDI is of some importance in all situations where compliance with dose constraints (see chapter 6) is tested. Some of the existing dose recommendations and limitations only refer to the local dose from a single slice rather than the overall radiation exposure. Therefore it seems reasonable and legitimate in those cases to correct for pitch-related effects by using the effective
Chapter 2: Fundamentals of CT Dosimetry

CTDI to demonstrate compliance with legal requirements.

However, the problem of increasing confusion remains, because yet another quantity has been introduced in a situation where the number of relevant dose quantities might better be reduced. In addition, it is not necessarily obvious to the user of the scanner that a correction for pitch-related effects has already been made, because the same term (‘CTDI₇₀’) is used. Therefore special care is necessary when utilizing such effective CTDI values for additional dose calculations, in order not to apply corrections for pitch-related effects a second time. The same holds true for some types of scanner which already correct the mAs displayed according to the pitch.

Dose Free-in-Air on the Axis of Rotation

Dose free-in-air on the axis of rotation, simply named CTDIₜₐᵢʳ, has up to now been the most important dose quantity for CT in Germany. It is also used more widely, however, for calculations of organ doses and effective dose (see later in this chapter) since all conversion tables require dose free-in-air on the axis of rotation as the input parameter. This quantity also represents a kind of CTDI, because it is obtained in a similar way as the central CTDI₁₀₀,c value, i.e. by measuring the dose-length product of the dose profile for subsequent division by the nominal slice thickness, h:

\[
CTDI_{tair} = \frac{1}{h} \int_{-50}^{50} K_{tair}(z) \cdot dz \quad (2.10)
\]

The only, but essential difference to other types of CTDI is that the CTDIₜₐᵢʳ is determined without using any phantom (hence the term ‘free-in-air’). The dose quantity is air kerma Kₜₐᵢʳ; dose values are stated in mGy.

Dose free-in-air on the axis of rotation was chosen as the only CT-related dose quantity in German standards and legislation because all the performance tests on installed scanners (acceptance and constancy tests) could then be performed without the need of bulky (and expensive) Perspex phantoms. Due to its restricted importance only for the German market, however, this quantity was never stated in the official specification sheets from CT manufacturers. Therefore CTDIₜₐᵢᵢ values were only available locally as a result of acceptance testing.

So-called ‘phantom factors’ P, which allow the conversion from CTDIₜₐᵢᵢ to CTDIₜ (and vice versa), depend on scanner-specific parameters, such as focus-to-axis distance, beam filtration, beam shaper and the diameter of the phantom to which CTDIₜ refers. The factors usually differ from one scanner model to another and are defined as

\[
P_H = \frac{CTDI_{tair,H}}{CTDI_{tair}}, \quad P_B = \frac{CTDI_{tair,B}}{CTDI_{tair}} \quad (2.11)
\]

The suffix ‘H’ refers to the head phantom with a diameter of 16 cm, the suffix ‘B’ to the body phantom (diameter 32 cm). Phantom factors and corresponding CTDI values for a number of older and newer scanner models are given in the appendix.

Dose free-in-air on the axis of rotation has up to now been of great importance in Germany for acceptance testing (DIN90), for dose recommendations related to the use of scanners (BÄK92) and for dose limitation (RöV98). Furthermore, it is at present the only dose quantity that enables the use of conversion tables in order to calculate organ doses and effective dose (Jones91, Zankl91, Zankl93). The simplicity of measuring CTDIₜₐᵢᵢ without the need of a phantom is another important advantage of this quantity. All the limitations mentioned previously for CTDI as an indicator of patient dose also apply to CTDIₜₐᵢᵢ. There are a number of additional drawbacks, however, which make this quantity even more questionable:

- CTDIₜₐᵢᵢ is far from being representative of organ doses because no phantom is used to simulate, at least roughly, the body;
- dose limitation based on CTDIₜₐᵢᵢ will treat preferentially scanners using heavily filtered beams over those with less beam filtration;
- the important dose reduction effects from beam shapers are not properly taken into account, since measurements are made only on the axis of rotation where these shapers have least influence.

This is why CTDIₜₐᵢᵢ will largely be replaced by CTDIₜ in the near future (e.g. when reference dose values are set up). CTDIₜₐᵢᵢ will maintain its importance in the field of dose calculations (organ dose, effective dose), as long as conversion tables based on CTDIₜ are not available.

Organ Dose

Organ dose is defined as the energy that is absorbed in a particular organ of the body (e.g. thyroid, lungs etc.), divided by the mass of the organ:

\[
D_{org} = \frac{\text{Absorbed energy}}{\text{Organ mass}} \quad (2.12)
\]
Organ doses are always average values, since the absorbed energy is averaged over the total mass of the organ. This is of great importance whenever organs are only partially irradiated, as in the case of organs extending over the whole body (e.g. red bone marrow) or organs situated at the borders of the scan range. Therefore, organ dose is not a pure physical 'dose' quantity; organ dose is nearly equal to CTDI only for those organs completely located inside the scan range of an examination.

Measurements of organ doses require the use of body-equivalent, anthropomorphic phantoms (e.g. Alderson phantom) to represent the patient. Such measurements are made with thermoluminescent dosemeters (TLD). The dose quantity for organ doses is dose equivalent; dose values are reported in μSv or mSv.

Alternatively, organ doses can be calculated from tabulated conversion coefficients. In most cases, these factors result from computational modelling with mathematical phantoms that are regarded as being representative of an averaged-sized human body. Conversion coefficients are by no means constant; they vary with a number of parameters such as tube potential, beam filtration, scanner geometry, organ position etc. The conversion tables of Zankl et al. for adults (Zankl91) provide conversion coefficients for both sexes (using the phantoms 'EVA' and 'ADAM', respectively). The input parameter for such organ dose calculations is CTDI\textsubscript{air}.

The conversion coefficients tell us the contributions to dose for each particular organ as a fraction of the dose free-in-air for each irradiated slice.

Dose quantities appropriate for CT

Organ doses for an examination are obtained by summing up the contributions from all slices inside the scan range:

\[
D_{org} = \frac{1}{p} \cdot \text{CTDI}_{air} \sum_{z} f(\text{organ}, z) \quad (2.13)
\]

If the table feed and slice thickness are not numerically equal, a correction is applied according to the pitch factor. The meaning of the symbols used in equation (2.13) is: \(p\) = pitch, \(\text{CTDI}_{air}\) = dose free-in-air on the axis of rotation, \(f(\text{organ}, z)\) = organ-specific conversion coefficient for a specified location \(z\) on the axis of rotation, \(z^-\) and \(z^+\) = lower and upper boundaries of the scan range. Conversion coefficients for individual slices are tabulated in increments of 1 cm in (Zankl91), and only 0.5 cm in (Jones91).

The significance of organ dose with respect to radiation exposure is obvious: by multiplying the organ dose for a particular organ by the corresponding risk factor, the probability of radiation-induced cancer can be assessed. In those cases where pregnant women have undergone a CT examination of the abdomen or pelvis, considerations on whether the pregnancy should be terminated can be made on a quantitative basis. However, dealing with organ dose becomes somewhat difficult when there is more than one organ to be considered. This holds true for most X-ray procedures. An elegant solution to this dilemma is given by the quantity 'effective dose', which is discussed at the end of this section.

**Dose-Length Product (DLP)**

CTDI, weighted CTDI and dose free-in-air on the axis of rotation are by definition only indicators of the level of local dose in the irradiated slice. Much confusion can arise when raising the question of the dose from an examination consisting of e.g. 15 slices. Even the authors of well-known teaching books, like 'Laubenberger', have some problems in giving a correct (or at least an unequivocal) answer to this question. According to (Laubenber90), the dose from an examination with 8 slices is eight times the dose from a single slice (1 slice: 15 mGy; 8 slices: 120 mGy). Remarkably, in the latest 6th edition, this problem is not even considered anymore.

The correct answer - which may surprise many people - is that the dose is the same in both cases. This seems a contradiction to our intuition suggesting that the radiation exposure should become more significant when increasing the extent to which the body is irradiated. Of course, the real cause of this confusion is in the definition of dose itself: 'dose' is defined as the amount of absorbed energy, \(dE\), per mass element, \(dm\). The unit in which dose is stated is 'gray', which is equal to one joule per kilogram (J/kg). 'Dose' therefore doesn't represent the total amount of absorbed energy - in the same way that the density of a body (which is given in g/cm\(^3\)) is something different than its mass. When the number of slices is increased, the irradiated mass grows by the same amount as the energy absorbed. Therefore the 'dose' doesn't change.

To simplify: 'dose' is an indicator of the intensity of irradiation inside the irradiated part of the body - but only there. If an organ, such as the liver, is already completely situated inside the scan range, then the dose to the liver remains the same even if the scan range is further extended (apart from some additional contributions from scattered radiation that are here neglected for simplicity). Only in those cases where an organ was partially irradiated, will the organ dose grow with increasing number of slices, but only once again until the organ
is fully irradiated. Therefore it is obviously erroneous to assume a proportionality between the ‘dose’ and the number of slices.

The same problem appears whenever a body is only partially irradiated; in diagnostic radiology, this is the standard situation. In conventional projection radiography, the quantity ‘dose-area product (DAP)’ is used to express both aspects of an irradiation - intensity and extent. The analogy for CT - where the diameter of the body is always completely irradiated - is the dose-length product (DLP). DLP is obtained by multiplying one of the dose quantities appropriate for CT (CTDI<sub>w</sub> or CTDI<sub>air</sub>) with the product of the number of slices n and the slice thickness h (in cm!):

\[
DLP_{XYZ} = CTDI_{XYZ} \cdot n \cdot h \quad (2.14)
\]

The unit in which dose-length product is stated is ‘mGy·cm’. The suffix (‘w’ or ‘air’) tells us which kind of CTDI was used. This difference is important if DLP is used to assess effective dose (see next chapter). In this definition of DLP, pitch-related effects have already been taken into account, since instead of the scan length L, the product of the number of slices and their thickness is used.

Since the pitch factor is already implicitly contained in equation (2.14), some care is necessary in order not to correct for pitch for a second time when calculating DLP from the effective CTDI (or the effective mAs value) displayed at the operator’s console. In these cases, the calculation of DLP<sub>w</sub> must be made using a different formula such as

\[
DLP_{w} = CTDI_{w,\text{eff}} \cdot p \cdot n \cdot h \quad (2.15)
\]

so as to convert from effective CTDI into weighted CTDI. As the product p · h is equal to table feed TF (see equation (2.1)), DLP<sub>w</sub> can also be calculated via

\[
DLP_{w} = CTDI_{w,\text{eff}} \cdot n \cdot TF \quad (2.16)
\]

where the product n · TF is equal to the scan length L.

It would appear that the dose-length product discussed here is the same quantity as that already mentioned in the context of the definition of CTDI, i.e. the area of the dose profile. It is important, however, to recognize that here the DLP stands for the entire scan series, with ‘length’ meaning the dimension of the irradiated part of the body. In case of the dose-length product for a single slice, however, ‘length’ means the dimension over which

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**Table 2.3**

*Overview of the dose quantities appropriate for CT which have been defined and discussed in this chapter.*

<table>
<thead>
<tr>
<th>Quantity</th>
<th>Symbol</th>
<th>Unit</th>
<th>Equation</th>
<th>Characteristics</th>
<th>Remarks</th>
</tr>
</thead>
<tbody>
<tr>
<td>Computed Tomography Dose Index</td>
<td>CTDI, mGy</td>
<td>(2.2)</td>
<td>Local dose</td>
<td>General dose descriptor for CT; coarse estimate of organ dose.</td>
<td></td>
</tr>
<tr>
<td>Multiple Scan Average Dose</td>
<td>MSAD, mGy</td>
<td>(2.3)</td>
<td>Average local dose</td>
<td>As CTDI but corrected for pitch factor.</td>
<td></td>
</tr>
<tr>
<td>CTDI (FDA)</td>
<td>CTDI&lt;sub&gt;FDA&lt;/sub&gt;, mGy</td>
<td>(2.4)</td>
<td>Local dose</td>
<td>Outdated; underestimates dose of thin slices.</td>
<td></td>
</tr>
<tr>
<td>CTDI (100)</td>
<td>CTDI&lt;sub&gt;100&lt;/sub&gt;, mGy</td>
<td>(2.5)</td>
<td>Local dose</td>
<td>Current definition of CTDI.</td>
<td></td>
</tr>
<tr>
<td>Weighted CTDI</td>
<td>CTDI&lt;sub&gt;w&lt;/sub&gt;, mGy</td>
<td>(2.8)</td>
<td>Average local dose</td>
<td>Main descriptor of local dose.</td>
<td></td>
</tr>
<tr>
<td>Effective CTDI</td>
<td>CTDI&lt;sub&gt;eff&lt;/sub&gt;, mGy</td>
<td>(2.9)</td>
<td>Average local dose</td>
<td>As CTDI&lt;sub&gt;w&lt;/sub&gt; but corrected for pitch factor.</td>
<td></td>
</tr>
<tr>
<td>Dose free-in-air on the axis of rotation</td>
<td>CTDI&lt;sub&gt;air&lt;/sub&gt;, mGy</td>
<td>(2.10)</td>
<td>Local dose</td>
<td>Easy to measure, but not very relevant; important input parameter, however, to calculate organ dose and effective dose.</td>
<td></td>
</tr>
<tr>
<td>Normalized CTDI</td>
<td>nCTDI&lt;sub&gt;XYZ&lt;/sub&gt;, mGy/mAs</td>
<td>(2.6)</td>
<td>Output</td>
<td>Characteristic quantity of a scanner.</td>
<td></td>
</tr>
<tr>
<td>Organ dose</td>
<td>D&lt;sub&gt;org&lt;/sub&gt;, mSv</td>
<td>(2.12)</td>
<td>Average dose in an organ</td>
<td>Usually calculated with tabulated conversion coefficients.</td>
<td></td>
</tr>
<tr>
<td>Dose-length product</td>
<td>DLP&lt;sub&gt;XYZ&lt;/sub&gt;, mGy·cm</td>
<td>(2.14)</td>
<td>Radiation exposure</td>
<td>Main descriptor of integral dose.</td>
<td></td>
</tr>
<tr>
<td>Effective dose</td>
<td>E, mSv</td>
<td>(2.17)</td>
<td>Radiation exposure</td>
<td>Allows dose comparisons with conventional X-ray examinations.</td>
<td></td>
</tr>
</tbody>
</table>
the contributions from the dose profile are summed. To avoid confusion, the dose-length product for a scan series will be abbreviated as ‘DLP’, whereas the dose-length product of the dose profile used for the measurement of CTDI is named ‘dlp’.

In complex examinations such as CT angiography or multi-phase examinations, more than one series of scans is subsequently made. Dose-relevant parameters such as mAs product, slice thickness and scan length may differ from series to series. Therefore the DLP should be calculated separately for each scan series. The total radiation exposure for the complete examination is obtained by simply adding the contributions from each series.

Future dose recommendations will be made in terms of both weighted CTDI and DLP (based on CTDI\textsubscript{w}). Thus attention will be drawn not only to the intensity of irradiation, but also to its geometrical extent. Therefore DLP plays an important role as an indicator of radiation exposure of the patient.

**Effective Dose**

All the dose quantities mentioned up to now (with the exception of organ dose) do not allow any comparison with dose values reported for conventional projection radiography. The only dose quantity that enables such a comparison is effective dose. With effective dose, the organ doses from a partial irradiation of the body are converted into an equivalent uniform dose to the entire body.

Effective dose $E$ (ICRP91) is defined as the weighted average of organ dose values $D_{org,i}$ for a number of specified organs:

$$E = \sum w_i \cdot D_{org,i}$$

(2.17)

How much a particular organ contributes to effective dose depends on its relative sensitivity for radiation induced effects, as represented by the tissue weighting factor $w_i$ attributed to the organ. The sum of all tissue weighting factors $w_i$ is equal to 1. The following values for $w_i$ are given in (ICRP91):

- 0.20 for gonads;
- 0.12 for each of lungs, (lower) colon, red bone marrow and stomach wall;
- 0.05 for each of breast, urinary bladder, liver, thyroid and oesophagus;
- 0.01 for each of skeleton and skin;
- 0.05 for the ‘remainder’.

The ‘remainder’ consists of a group of additional organs and tissues with a lower sensitivity for radiation induced effects for which the average dose must be used: (upper) colon, small intestine, brain, spleen, muscle tissue, adrenals, kidneys, pancreas, thymus and uterus.

If a single one of the remainder organs receives a higher dose than any of the 12 organs with specific weighting factors, then the dose to that particular ‘remainder’ organ is weighted by a factor of 0.025; in this case, the average dose to the other organs in the remainder group is weighted by a factor of only 0.025. This scenario is of particular importance for head examinations.

Effective dose cannot as such be measured. It can be assessed in various ways, however, from knowledge of the dose free-in-air on the axis of rotation or the dose-length product. Appropriate methods will be described in the following chapter.

**Summary of CT Dose Quantities**

In table 2.3, an overview is given of the dose quantities appropriate for CT which have been defined and discussed in this chapter. In summary, it is most important to distinguish between (local) dose and (integral) radiation exposure; to not use the outdated and incorrect quantity CTDI\textsubscript{FDA}; to not mix up CTDI (dose) and normalized CTDI (output); and to use dose free-in-air, not for any comparisons, but exclusively as an input parameter for calculations of organ dose and effective dose.