METHODS FOR IN VIVO DOSIMETRY
IN EXTERNAL RADIOTHERAPY
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METHODS FOR IN VIVO DOSIMETRY
IN EXTERNAL RADIOTHERAPY

J. Van Dam and G. Marinello

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I. INTRODUCTION

Although the relevance of quality control is accepted, its practical realization is not however straightforward due to the complexity and the many steps involved in the radiotherapeutic process. Some physics steps such as basic dosimetry and the mechanical status of the equipment are, due to modern dosimetry protocols, subject to clearly written quality assurance “guidelines”, while for other steps of radiotherapy the situation is not so clear. For example there is a lack of systematic quality control at the level of the treatment planning systems, the patient set-up and the delivery of the correct number of monitor units. Further systematisation of the quality control for these areas, and others of equal importance in the treatment chain, requires in many parts of the world additional radiotherapists, medical physicists, and technicians. The situation is however slowly improving at the national, regional, and wider international levels due to the efforts of a few highly motivated individuals.

Another approach, which complements step-by-step quality control rather than competes with it, is to judge the end-product, which is the treatment as delivered to the patient, and to try to “trace back” any observed deviations to the faulty step involved (Van Esch et al 2002; Dempsey et al 2000).

Port films belong to the quality control arsenal for checking geometrical parameters: their evaluation allows to check whether the treatment volume is covered and whether the healthy tissues are spared adequately. Further improvement in this field, mainly through the increasing availability of digital techniques, is at present a matter of extensive research.

While port films are routinely used to assess the volumes which have been treated in the patient (but not the dose distributions), it is also important to check the absorbed dose which has been delivered in practice. This can be done by putting dosimeters on the patient’s skin or in natural cavities. This method is called “IN VIVO DOSIMETRY” and its methodology is the subject of this work. It is usually performed to detect errors in individual patients (Bascuas et al, 1977; Lahtinen et al, 1988; Marcie et al, 1991-a and b; SFPH and SFRO, 1992; Mijnheer, 1994; Leunens et al, 1994; van Bree et al, 1994; Van Esch et al, 2002; Ciocca et al. 2003; Higgins et al, 2003), to detect errors in core procedures (Leunens et al, 1990-a and b; Hazle et al, 1992), to evaluate the quality of specific treatment techniques (Marinello et al, 1980 and 1982; Meeting of Leiden, 1982; AAPM, 1986; AAPM, 1987) or to evaluate the dose in situations in which the dose calculation is inaccurate or not possible (Rudên, 1976; Rosenblum, 1982; Rosenbloom et al, 1982; Burleson et al, 1991; Butson et al 1998-b).

This work is an attempt to provide information concerning the methodology to be recommended for high energy photon and electron beams according to the medical situation as well as the specific methods to be used in this area (Chapter II). Moreover it indicates the typical characteristics of the detectors which can be used in in vivo dosimetry such as diodes (Chapter III) and radiothermoluminescent dosimeters (Chapter IV). Advantages and disadvantages of both of them are discussed in chapter V. The last chapter VI summarizes the
alternative methods which are already in use or could present an interest for in vivo dosimetry in a near future. The methods indicated are intended to obtain an accuracy of the order of one to two percent in the doses determined. Use of these methods for in vivo dosimetry in IMRT falls outside the scope of present work and is thus not addressed.

Due to the highly technical aspects of some chapters, some examples and practical consequences illustrating how to make use of the subject covered have been included. Theoretical aspects are not treated in this work, but the reader could refer to the extensive list of references added at the end of this booklet.
II. CLINICAL APPLICATIONS AND ASSOCIATED METHODOLOGY

A first possible aim of \textit{in vivo} dosimetry is to compare the doses derived from the signal of the detectors placed on the skin with the theoretical values, as calculated by the Treatment Planning System (TPS). As however the accuracy of the calculation of the dose to the skin is questionable, and in many cases irrelevant, the signal of the detector is converted to the dose, at a point which is still close to the skin, but at a certain depth where the accuracy of the TPS is much more satisfactory. One point is close to the entrance, while the other is close to the exit surface of the beam. The corresponding doses are called entrance and exit doses, respectively. They will be defined in §II.1. With regard to the exit dose, one should realize that in the real patient there is in most cases a considerable loss of backscatter, while the TPS calculations are valid for semi-infinite patients implying complete backscatter at the exit surface. A correction is then necessary (Rizzotti et al, 1985; Knöös et al, 1986; Leunens et al, 1990-b).

A more ambitious aim of \textit{in vivo} dosimetry is to check the target dose (ICRU, 1993, 1999 and 2004), in order to verify the correct delivery of irradiation. Except when detectors can be introduced in natural body cavities such as oesophageal tube, rectum, vagina, etc, this is impossible. As a matter of fact, a check of the entrance and exit dose, is also an indirect check of the target dose. However, if a deviation is observed between the computed and measured entrance or exit dose (under the assumption that the experimental value is correct) it may be because the target dose is wrong (due to a wrong number of monitor units, an error in the irradiation parameters, an incorrect patient set-up or an unexpected variation of the machine output), because the calculation of the entrance or exit doses, even from a correct target dose, is wrong, or because of a combination of both types of error. A more selective check of the target dose is then of high interest.

A third possible aim of \textit{in vivo} dosimetry can be the determination of the skin dose itself. This measurement is critical and requires a special methodology (Karzmark, 1968; Marshall and Docherty, 1971; Marinello et al, 1977 and 1980; Contento et al, 1984; AAPM, 1987; Butson et al 1998-b).

High energy photon beams and electron beams will be considered separately in this chapter. Total body irradiations using these two different types of beams will be also considered.

II.1. High energy photons

In the case of a single beam the entrance dose $D_{\text{entrance}}$ and the exit dose $D_{\text{exit}}$ are defined at points at a certain distance from the patient surface at the entrance and exit of the beam (referred to as the “entrance” and “exit surface of the patient”, respectively). This distance is equal to the “depth of maximum build-up” $d_{\text{max}}$ (Figure II.1). This definition of
D_{exit}, symmetrical to D_{entrance} with respect to the midline, is a simplification in the frequent use of two opposed beams, and is useful for the derivation of the target dose D_{target} (§ II.1.1.4.).

It is essential to check each beam contributing to the target dose individually, at least at the first treatment session, in order to identify the possible causes of errors and to correct them. In the particular case of TL dosimetry that implies the need to change the set of detectors after each irradiation beam.

Figure II.1: Schematic representation of the different doses involved in *in vivo* dosimetry for a single beam. The surface dose (D_{surface}) is defined at 0.05cm below the entrance surface, the entrance dose (D_{entrance}) at d_{max}, the target dose (D_{target}) at the depth of dose specification, the exit dose (D_{exit}) at a distance of d_{max} from the exit surface on the beam axis. The definition of exit dose implies conditions of complete electron backscatter (because d_{max} is larger than the electron backscatter range (s) but smaller than the photon backscatter range (r)).

Once the quality of the irradiation delivered individually by each beam has been checked at the first treatment session, some users may wish to check also the reproducibility of the treatment during the following sessions. In order to save time, they might then prefer to leave the same *in vivo* detectors on the patient for the full treatment session including all beams. As long as the entrance and exit doses of each beam are not influenced by contributions from other beams, which is still often well approximated in the case of convergent beams, the definitions of D_{entrance} and D_{exit} are still valid for each beam individually. However this is not always the case, and certainly not for 2 parallel opposed beams (Figure II.2) for which D_{entrance} and D_{exit} lose their meaning and should be replaced.
by the dose $D_{M,1}$ and $D_{M,2}$ at depths of maximum dose from the entrance side of each beam $d_{M,1}$ and $d_{M,2}$ respectively (§ II.1.2).

Figure II.2: Schematic representation of the different doses involved in in vivo dosimetry for 2 parallel opposed photon beams. The beams are equally weighted at the level of the target, which is either at midline with $d_1=d_2$ (A) or in an non-central position with $d_1<d_2$ (B). The entrance and exit doses shown in Figure II.1 are replaced by the dose $D_{M,1}$ at the depth of maximum dose $d_{M,1}$ at the entrance side of beam 1, and by the dose $D_{M,2}$ at $d_{M,2}$ at the entrance side of beam 2.

II.1.1 Single beam

II.1.1.1 Entrance dose

At the entrance side of a medium irradiated by a single photon beam the dose gradually increases from a low value at the surface up to a maximum value $D_{\text{entrance}}$ at a depth $d_{\text{max}}$ which depends upon the energy, the collimator opening, the skin-source-distance (SSD), the introduction of beam modifying devices and the distance separating them from the patient skin, etc. Generally the increase of dose as a function of depth from surface to $d_{\text{max}}$ is the steepest just below the surface, gets less pronounced at larger depths and finally levels off at $d_{\text{max}}$ (Figure II.1). That means that the measurement of $D_{\text{entrance}}$ must be carried out with enough material in front and around the detector placed at skin level in order to be reproducible.

Most of the detectors used for in vivo dosimetry have a sensitive part about 1 mm thick, or less, which means that, when they are on the skin, they integrate the dose in a region of very steep dose gradient. This complicates greatly an accurate determination of the ratio
between dose to the detector and $D_{\text{entrance}}$. Moreover, when the bare detector is used it is subject to almost the full headscatter contaminating electron spectrum. The number of these electrons is known to increase as a function of the collimator opening and to decrease as a function of SSD. In order to limit as much as possible the influence of headscatter electrons on $D_{\text{entrance}}$, it is necessary to use a build up cap the dimensions of which correspond to the dimensions necessary to ensure full build up for the smallest collimator opening in the absence of any accessory (Figure II.3 and Table II.1). The problem is that with the higher energies the thickness of the build-up cap can be such (several centimeters of tissue-equivalent material) that it might compromise patient comfort and lead to an underdosage of the treatment volume, combined with loss of skin sparing, in a large area (Figure II.4).

A way to reduce the build-up cap dimensions is to use a high density material, but in this case the build-up cap can change the response versus energy of the detector surrounded by it. All the precautions should be taken to avoid errors and uncertainties: e.g. calibration should be performed for the detector together with the build-up cap. As in this case the dimensions of the build-up cap are small, a possible way to decrease the disturbing effect mentioned above, is to introduce small variations in the daily position of the detector in order to smear out the dose perturbation. This is however only possible when the lateral dose distribution is sufficiently homogeneous, and may not be very convenient in practice.

![Figure II.3: Variation of the response of diodes as a function of collimator opening (Scanditronix EDP-20 and EDE for 18 MV and $^{60}$Co, respectively). When the detector is covered with a full build-up thickness, a variation less than 1% is observed either for cobalt 60 or for 18 MV X-rays, when increasing collimator opening from 5 cm x 5 cm to 30 cm x 30 cm. By contrast, when applying only 2 cm build-up for 18 MV X-rays, the increasing contribution of headscatter electrons causes an apparent sensitivity and a more pronounced increase as a function of collimator opening is observed.](image-url)
Table II.1: Examples of maximum build-up \( d_{\text{max}} \) for photon beams as a function of energy and field size. These data give an indication of the maximum thickness for the build-up cap to be used on the detector.

<table>
<thead>
<tr>
<th>Photon energy</th>
<th>( d_{\text{max}} ) Field 5 cm x 5 cm</th>
<th>( d_{\text{max}} ) Field 30 cm x 30 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>cobalt-60</td>
<td>0.4 cm</td>
<td>0.2 cm</td>
</tr>
<tr>
<td>4 MV</td>
<td>1.0 cm</td>
<td>0.8 cm</td>
</tr>
<tr>
<td>6 MV</td>
<td>1.7 cm</td>
<td>1.5 cm</td>
</tr>
<tr>
<td>18 MV</td>
<td>3.5 cm</td>
<td>2.0 cm</td>
</tr>
<tr>
<td>25 MV</td>
<td>4.4 cm</td>
<td>2.5 cm</td>
</tr>
</tbody>
</table>

Figure II.4: Dose profiles in a polystyrene phantom when a detector is placed on its surface. The profiles were measured at surface, \( d_{\text{max}} \) and 5 cm depth. For 4 and 21 MV X-rays the detector was covered with material with a tissue-equivalent thickness of 8 and 20 mm respectively (After Nilsson et al, 1988).
In order to achieve most of the electronic equilibrium and yet to allow for better and easier handling, some users propose to decrease the thickness and the lateral dimensions of the cap. Accurate measurements being only possible if the detector is not in too high a dose gradient, the build-up thickness is a compromise between minimum field disturbance and minimum dose gradient over the detector. Moreover in case of incomplete build-up, some headscatter electrons will still reach the detector (Figure II.3). Attention should be paid in this case to the need of correction factors established on a phantom in the same irradiation conditions as for patient (§ III.4.1 and IV.4.1). They generally depend upon the geometry of the irradiation (field size, SSD, etc.) and the presence or not of accessories on the treatment head.

It is important to keep in mind that incomplete build-up on an entrance detector might lead to certain correction factors which are only determined by the influence of the headscatter electron contamination and not at all by the intrinsic characteristics of the detector type involved. Those latter category of correction factors are treated in the chapters about diode dosimeters (chapter III) and radiothermoluminescent dosimeters (chapter IV).

II.1.1.2. Exit dose

At the exit side of the patient there is a build-down region related to lack of backscatter radiation from the air behind the patient (Figure II.1). This lack of backscatter concerns photons as well as secondary electrons. While the lack of electron backscatter causes a build-down of the dose only in the latter few millimetres in front of the exit surface of the patient (approximately from 1 mm for cobalt 60 to 3 mm for 20 MV X-rays (Lambert et al, 1983), the lack of photon backscatter influences a much deeper region and increases as a function of field size. Moreover the extension of the region where an increase of the thickness of the backscatter layer results in a significant increase of the exit dose has been observed to decrease as a function of photon energy: for a 10 cm x 10 cm field, it extends to more than 10 cm from the surface for cobalt-60 gamma-rays but it is only about 1 cm for 20 MV X-rays (Chavaudra et al, 1973; Gagnon and Orton, 1979).

Due to this two-component backscatter near the surface of the beam, the position at which the exit dose $D_{\text{exit}}$ should be defined is much less obvious than for $D_{\text{entrance}}$. Mainly for purpose of derivation of the target dose (§ II.1.1.4), in this work $D_{\text{exit}}$ is taken at $d_{\text{max}}$ from the exit surface, e.g. symmetrically with the entrance dose with respect to the midline. In the real patient, at $d_{\text{max}}$ from the exit surface, the electron backscatter is complete, but in most conditions only partial photon backscatter is achieved (Knöös et al, 1986).

To derive the exit dose $D_{\text{exit}}$ at $d_{\text{max}}$ from the exit detector signal (Figure II.1), it is necessary to cover the detector with enough material behind and around it in order to ensure complete electron backscatter (otherwise it would be in too high a dose gradient, which would decrease the accuracy). Although thicknesses of build-up and build-down regions are different, most users are accustomed to use in practice the same caps for in vivo
measurements performed on patient treated with opposed fields. The ratio between the dose to the detector and the exit dose has again to be determined in phantom in the same irradiation conditions as for the patient.

Mostly the \textit{in vivo} measurement of the exit dose is performed concomitantly with that of the entrance dose. It is then important when positioning the exit detector, to avoid the shadowing effect of the entrance detector.

\textbf{II.1.1.3. Surface dose}

To measure the dose to the skin which is defined at 0.05 cm under the surface (ICRU, 1984) and is called the surface dose $D_{\text{surface}}$ (Figures II.1 and II.2), thin detectors must be used. When they are very thin, such as monocoated photographic emulsions (AAPM, 1986), they should be covered with about 0.05 cm of build-up material. When thicker detectors are used, such as some TL chips or thin layers of TL powder (Chapter IV) wrapped in envelopes made of paper, they have to be stuck on the skin without any build-up material. Correction factors have to be applied to their response when their effective point of measurement is not at 0.5 mm under their surface (Kron et al, 1993-a and b).

\textbf{II.1.1.4. Target dose}

\textbf{II.1.1.4.1. General Methodology}

A very simplified approach consists of considering the target dose $D_{\text{target}}$ equal to the mean of $D_{\text{entrance}}$ and $D_{\text{exit}}$. This method, which can be acceptable in some practical conditions (see example), may induce errors of several percents in others.

A more accurate method, published by Rizotti et al (1985) and Leunens et al. (1990-b), is based on the symmetrical (with respect to the midline point) expansion or compression of the real patient to a thicker or thinner “water patient”. However a prerequisite is that tissue inhomogeneities should be symmetrical and equally distributed with respect to the midline for reliable target dose determination. Most regions of the human body are well suited for this type of calculation in left-right direction, because of the symmetrical disposition of the different types of tissues irradiated (Figure II.5.A). Unfortunately, except for the brain and skull, the method is not straightforwardly applicable in antero-posterior direction (Figure II.5.B), because the symmetry is not respected.

The method is based on the concept of exit transmission $T_{\text{exit}}$ and midline transmission $T_{\text{mid}}$, which are defined as the ratio of the dose at depths of $(z-d_{\text{max}})$ or $z/2$ respectively and the entrance dose (Figure II.6). The depth $z$ is the water equivalent depth of the patient. Then the estimation of the target absorbed dose consists of three steps:
- firstly, the determination of the measured exit transmission ($T_M$) for a given patient:
  $$T_M = \frac{D_{M,\text{exit}}}{D_{M,\text{entrance}}}$$
- secondly, the derivation of the midline transmission ($T_{\text{mid}}$) for the given patient from the measured exit transmission with the help of two groups of curves (§ II.1.1.4.2)
- finally, the estimation of the absorbed midline dose ($D_{\text{mid}}$) from the product of the measured entrance dose ($D_{\text{M,entrance}}$) and the midline transmission ($T_{\text{mid}}$).

Figure II.5: Examples of symmetrical (A) and asymmetrical (B) disposition of the tissues, with respect to the midline depth. For the head-and-neck fields (A) the method of target dose determination is applicable, while it is not for the thorax fields (B).

Figure II.6: Schematic representation of the doses used in the definition of the different transmissions. The exit and midline transmissions, $T_{\text{exit}}$ and $T_{\text{mid}}$ respectively, are defined respectively as the ratio of the exit dose $D_{\text{exit}}$ and the dose $D_{\text{mid}}$ at midline depth $z/2$, to the entrance dose $D_{\text{entrance}}$. 

10
II.1.1.4.2. Derivation of midline transmission

The exit and midline transmission curves needed for the derivation of the midline transmission from the measured exit transmission are theoretical curves calculated from tissue phantom ratio’s (TPR) taken at the level of either the exit of the beam or of the midline depth. The middle of the patient being assumed to be at the isocentre, at the source-axis-distance SAD from the source, one can write (under the assumption that the water-equivalent thickness \( z \) is not too different from the geometrical thickness, which condition is mostly fulfilled):

\[
T_{\text{exit}} = \frac{\text{TPR}(A', z - d_{\text{max}})}{\text{TPR}(A', d_{\text{max}})} \cdot \left[ \frac{\text{SAD} - \frac{z}{2} + d_{\text{max}}}{\text{SAD} + \frac{z}{2} - d_{\text{max}}} \right]^2 \cdot \frac{B_{A'}}{B_{A'}} \cdot \frac{1}{B_{A'}} \quad [1]
\]

\[
T_{\text{mid}} = \frac{\text{TPR}(4, z/2)}{\text{TPR}(4, d_{\text{max}})} \cdot \left[ \frac{\text{SAD} - \frac{z}{2} + d_{\text{max}}}{\text{SAD}} \right]^2 \cdot \frac{B_A}{B_{A'}} \quad [2]
\]

The TPR’s are corrected for inverse square law taking into account the differences in distances of the points compared. \( B_A, B_{A'}, \) and \( B_{A0} \) are the backscatter factors for the field sizes \( A, A' \) and \( A_0 \), respectively at the isocentre \((z/2)\), at the exit dose point situated at \((z - d_{\text{max}})\) and at the entrance dose point situated at \(d_{\text{max}}\) (Figure II.6).

\[
A' = A \cdot \frac{\text{SAD} + \frac{z}{2} - d_{\text{max}}}{\text{SAD}} \quad [3]
\]

\[
A_0 = A \cdot \frac{\text{SAD} - \frac{z}{2} + d_{\text{max}}}{\text{SAD}} \quad [4]
\]

\( B_{A'} \) is the correction factor for incomplete photon backscatter at \(d_{\text{max}}\) from the exit surface of the beam (Leunens et al, 1990-b).

Midline and exit transmission curves have been published for cobalt 60 (Rizzotti et al, 1985) and 6 MV X-rays (Leunens et al, 1990-b). In order to give an indication for higher energies, curves giving \( T_{\text{mid}} \) and \( T_{\text{exit}} \) as a function of field size \( A \) and \( A' \) and depths are shown in Figure II.7 for 18 MV X-rays. However, in most clinical cases, just taking the arithmetic mean of entrance and exit dose may be a quite reasonable approximation. Moreover, every user should be aware that the data available may not be applied blindly and require some experimental checks in his own beam.
Figure II.7: Transmission curves obtained for 18 MV X-rays. The exit and midline transmission $T_{exit}$ and $T_{mid}$ are plotted as a function of water equivalent thicknesses $(z)$ and $(z/2)$, respectively. Both groups of curves have been calculated for different equivalent square fields (from 5 x 5 to 20 x 20 cm$^2$). $T_{exit}$ and $T_{mid}$ have been calculated from Tissue-Phantom-Ratios, taking into account the actual position of the calibration points and the lack of backscatter (Leunens et al, 1990-b).

**Example:**

A 16 cm thick patient is treated isocentrically (SAD = 100 cm) with a 18 MV X-ray beam. The field size $A$ at the isocentre is equal to 10 cm x 10 cm. The measured entrance and exit doses are found equal to 117 cGy and 78 cGy, respectively, from which values a measured exit transmission of 67 % is derived. The field size $A' = 10.5$ cm x 10.5 cm (Figure II.6 with $d_{\text{max}} = 3.5$ cm) and the total equivalent thickness $z$, deduced from Fig.II.7 (arrow 1 and the abscissa of arrow 2) is equal to 17.3 cm. This means that the real patient with a diameter of 16 cm was “expanded” to a “water equivalent” patient with a diameter of 17.3 cm. The midline position was however not changed. Once the water equivalent thickness of the patient is known, the midline transmission is determined as the read-out on the y-axis of the $T_{mid}$ curve for the 10 cm x 10 cm field size $A$. For $z$ is 17.3 cm, $T_{mid}$ is 83 % (arrow 2 and arrow 3), yielding a midline dose of 117 cGy x 0.83 = 97 cGy. This result differs by less than 1% from the arithmetic mean of entrance and exit dose.
II.1.2. Two parallel opposed beams.

In the case of 2 parallel opposed beams the detector signals are the sum of an entrance and an exit dose components. These 2 signals should be correlated (e.g. by phantom measurements) to the maximal doses \(D_{M,1}\) and \(D_{M,2}\) in the patient along the beam axis at depths \(d_{M,1}\) and \(d_{M,2}\) which does not correspond necessarily to the depth \(d_{\text{max}}\) of each beam considered separately (Figure II.2). The difference between \(d_{M,1}\) or \(d_{M,2}\) and \(d_{\text{max}}\) has however been shown to be smaller than 1 mm for 4 and 25 MV X-rays (field 20 x20 cm² at SSD 100 cm). A prerequisite for using this method is, first of all, that the calibration factor of the detector be the same for both entrance and exit positions, which is not always the case (§ III.4.1 and IV.4.1).

II.1.3. Total Body Irradiation

*In vivo* dosimetry is of particular relevance in case of Total Body Irradiation (TBI) before bone marrow transplantation for different reasons (Meeting of Leiden, 1982; AAPM, 1986; ESTRO, 1987): difficulties in calculation of the dose at different points in the patient, increased risk of patient movements due to the long duration of the treatment and, in single fraction regimen, the necessity to correct the dose before the end of the session. In this respect, the *in vivo* measurements are to be considered not only as an independent check, but rather as an integral part of the overall dosimetric approach for this particular treatment technique.

The task for in vivo dosimetry in the case of TBI is threefold: to determine the dose at the dose specification point, usually taken at mid-pelvis or mid-abdomen (Figure II.8), to estimate the homogeneity of the midline dose distribution at different loci in cranio-caudal direction, to monitor the dose at the level of organs at risk (lungs, liver, etc).

The discussion in this section is deliberately limited to the application to TBI conditions of the general principles concerning entrance, exit and target dose measurements outlined above. Important aspects such as calibration methods and choice of appropriate detector type will be included subsequently at the level of the more general discussions of these topics.

The measurement of the entrance dose in TBI conditions is performed in a situation where the superficial tissue layers belong to the target volume. Especially at high photon energies it may be necessary to add some beam spoiler in order to reduce \(d_{\text{max}}\) (Rosenbloom et al, 1982), although this is not always the case (Dutreix et al, 1982). As a consequence the skin, and thus the entrance *in vivo* detectors, are almost in full build up conditions, so that no additional build up cap has to cover the detectors. In case of TBI this is clearly an advantage, because often the detectors remain on the patient for the full dose, which, if a build-up cap had to be used, could cause a problem of target underdosage (§ II.1.1).
As far as the exit dose measurement is concerned, some set-up related problems may also occur in TBI. Indeed it is not uncommon, in order to obtain a field which is as large as possible, to position the patient very close to the floor or a wall of the treatment room. The walls especially, being made of high-Z material (e.g. barytes concrete), may produce significant electron and photon backscatter components on the patient’s skin (Van Dam et al, 1988). Since the electron component is of very limited penetration, if it is not stopped by the build-down cap on the exit detector, it would obviously be the origin of an overestimation of the value of the exit dose which is used for calculation of the target dose. It is then important to check, and if necessary to correct for, the influence of this backscatter electron radiation.

In order to fulfill the dosimetric needs in case of TBI, target doses, at midline depth, will have to be estimated from measured entrance and exit doses according to the methodology discussed (§II.1.1.4.). The general principle (Van Dyk et al, 1980; Marinello et al, 1982; AAPM, 1986) that all dosimetric quantities used for TBI dosimetry have to be measured (or at least checked) in TBI conditions should not be forgotten: it is also valid for the TPR’s and backscatter factors used in formula [1] and [2].

II.2. High energy electrons

High energy electron beams are characterized by fast changes in the energy spectrum along the central axis and in the angular distribution of the electrons with patient depth (ICRU, 1984 and 2004). They are also characterized by a build-up less pronounced than for
photons even for accelerators using a few discrete beam limiting plates instead of cones as electron collimation (Table II.2). As the depth of penetration of electrons is limited, the notion of exit dose has no sense except for very thin anatomical structures such as fingers, hands, etc. As these structures are rarely involved in the electron fields used in clinical practice, except e.g. in the case of total skin irradiation, exit dose will not be considered in this work.

Table II.2: Dose variation in the build-up region for a tissue-equivalent material irradiated at variable energy levels in the electron beam of an accelerator type Sagittaire-CGR MeV (field size 10 cm x 10 cm at the isocenter situated at 105 cm from the source). The unit is equipped with an additional collimator. The doses, measured using a NPL plane parallel chamber, are expressed as a function of the maximum dose on the beam axis (After Marinello et al, 1981).

<table>
<thead>
<tr>
<th>Depth (“cm water”)</th>
<th>Electron energies (MeV)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>7</td>
</tr>
<tr>
<td>0</td>
<td>70.6</td>
</tr>
<tr>
<td>0.1</td>
<td>75.3</td>
</tr>
<tr>
<td>0.5</td>
<td>81.9</td>
</tr>
<tr>
<td>1.0</td>
<td>90.0</td>
</tr>
<tr>
<td>1.9</td>
<td>100 %</td>
</tr>
<tr>
<td>2.5</td>
<td></td>
</tr>
<tr>
<td>3.3</td>
<td></td>
</tr>
<tr>
<td>3.5</td>
<td></td>
</tr>
<tr>
<td>4.0</td>
<td></td>
</tr>
</tbody>
</table>

Most often in vivo dosimetry is based on measurements performed at the patient skin to be correlated with the target dose $D_{\text{target}}$ if necessary. Clinical applications are generally limited to single beams either used alone, or joined at the patient skin in order to cover a large surface (breast irradiation, total skin irradiation).

When measurements are performed in air cavities or behind bone structures (for instance in the mouth) problems of over- or underdosage can appear when treating with electron beams because of the inhomogeneities. In these cases in vivo dosimeters should be used with extreme care.

**II.2.1. Surface dose**

The dose to the skin is defined at 0.5 mm under the surface (ICRU, 1985) and is called the surface dose $D_{\text{surface}}$ (Figure II.9). It can be measured with thin detectors such as TL chips or thin layers of TL powder (Chapter IV) wrapped in envelopes made of thin paper stuck on the skin without any build-up material. When thicker detectors are used correction factors have to be applied to their response because their effective point of measurement is more than 0.5 mm under the surface (Kron et al, 1993-a and b). On the contrary when very
thin detectors such as monocoated photographic emulsions are used, they are to be covered with about 0.5 mm of build-up material. In any case it is necessary to make sure that the air space between the patient skin and dosimeter wrapping envelope be totally eliminated.

![Figure II.9: Schematic representation of the different doses involved for a single electron beam. The target dose \( D_{\text{target}} \) is the same as the entrance dose \( D_{\text{entrance}} \). Doses are defined on the beam axis.](image)

II.2.2. Target dose

For electron beams the target dose \( D_{\text{target}} \) is generally prescribed at \( d_{\text{max}} \) (ICRU, 1978 and 2004) and is therefore the same as the entrance dose \( D_{\text{entrance}} \) (Figure II.9). As \( d_{\text{max}} \) may be reached at depths greater than 1 cm (e.g. at depths varying from 1.9 to 4 cm for energies varying from 7 to 19 MeV as shown in Table II.2) the use of some build-up material may seem useful. However, because of the high scattering properties of the electrons, such a method can induce a relatively important scattering artefact and may alter the dose distribution substantially. For this reason it is necessary to derive the target dose \( D_{\text{target}} \) from \( D_{\text{surface}} \) using correction factors established on a phantom, in the same irradiation conditions as for the patient. These factors can be determined using a flat parallel chamber (without polarity effect) allowing an accurate determination of \( d_{\text{max}} \). Then detectors which are intended to be used for in vivo dosimetry are placed at the surface and at \( d_{\text{max}} \). The correction factor to be used is the ratio of both signals.

Because of the less pronounced build-up observed with electrons than with photons (Table III.2) the value of the correction factors can be obtained with an accuracy better than 5% if the detector equivalent-thickness is correctly taken into consideration.
In vivo dosimetry is of particular relevance in case of Total Skin Irradiation (TSI) because the dose distribution over the patient’s skin is heavily dependent upon the specific technique chosen, the particular equipment on which the treatment is carried out and the facility where it has been implanted. Irrespective of the technique used (Karzmark et al, 1960; Page et al, 1970; Marinello et al, 1977; AAPM, 1987) the penetration of the electron beams is limited to the first millimeters of the skin and the depth of maximum dose $d_{\text{max}}$ is generally situated very close to the surface. Dose uniformity is difficult to reach because of the varying curvature of body surface over the field. Moreover some parts of the patient can shield other parts (patient self-shielding). As a consequence the dose distributions vary widely and no simple generalization from one patient to another is applicable. The only solution to make sure that the prescribed dose is correctly delivered consists of carrying out in vivo measurements during patient irradiation to check the uniformity and identify areas requiring local complementary fields. Such measurements allow one to check also the reproducibility between the different treatment sessions.

Considering the low energy of electron beams to be used, the number of sites to be explored simultaneously (several tens) and the curvature of the body surfaces, radiothermoluminescent detectors of small thickness are the best suited to perform in vivo measurements in TSI conditions. They will be wrapped in envelopes made of thin paper and stuck on the skin without any build-up material. The methodology previously described in § II.2.2 will be used to calibrate them in the same irradiation conditions as those used for TSI and to derive $D_{\text{target}}$ from $D_{\text{surface}}$. Generally both these doses differ only by a few per cent. For some particular curvatures of surface anatomy they may even become identical.
III. DIODE DOSIMETERS

Diode dosimeters, which belong to the category of semiconductor detectors, are comparable to ionisation chambers but are generally used without external bias voltage and are more sensitive for the same detection volume. For *in vivo* dosimetry they provide the advantage of immediate availability of the signal but must be used with care.

III.1. Principle of the method

III.1.1. Semiconducting phenomenon

Semiconducting crystals, e.g. germanium or silicon, in their ground state are to be considered as electrical insulators. But only a small amount of energy input by raising the temperature is necessary to induce some conductivity in the material. The energy is transferred to electrons which may then quit the crystal lattice, creating a positive ‘hole’ due to the resultant charge defect. Not only the free electron, but also this hole will participate in the electrical conductivity. Indeed by re-arrangement of the electronic crystal structure, the hole will migrate through the crystal lattice and produce some net charge displacement. However, “electrons” and “holes” may be lost for the phenomenon of conductivity through recombination, which may take place at the site of imperfections in the crystal lattice.

The intrinsic semiconductivity can be increased considerably by doping the crystal of the semiconductor with some impurities. These impurities may be of either the “donor” or the “acceptor” type. Donor impurities add electrons to the crystal, while acceptor impurities take away and fix electrons from the crystal lattice where positive holes are created. Typical donors are phosphorous or arsenic, and typical acceptors are boron or aluminium. Semiconductors doped with a donor or an acceptor are of the n-type or p-type, respectively. The conductivity takes place mainly through electrons in an n-type semiconductor while in a p-type semiconductor mainly holes contribute to it. Electrons are referred to as minority carriers in a p-type and as majority in an n-type semiconductor.

III.1.2. p-n junction or diode as radiation detector

A p-n junction or diode is an internal boundary between p-type and n-type regions in a single crystal. The crystal is doped in two steps: e.g. if the diode is to be of the p-type, acceptor impurities are first added to the crystal, and then donor atoms (but in a much higher concentration than that of the acceptor) are diffused into the surface of the p-type material. At the transition from p- to n-type material, a charge-free “depletion layer” is formed, over which an electrostatic potential difference is created (about 0.7 V for a silicon diode). As a result an electric field $E$ is created over the depletion layer (Figure III.1). By
contrast to most other applications in electronics, in dosimetry no external bias is applied to the diode, which is then used in the “unbiased” or “short-circuit” mode.

**Figure III.1**: Principle of diode detection in short-circuit mode. The radiation produces electrons and holes in the depletion layer, which are attracted by the positive and negative side of the diode, respectively. As a result a current I proportional to the number of charges created, will flow in an external circuit.

When the diode is irradiated, an ionisation “electron-hole” pair may be created in the depletion layer (Figure III.1). The electrons and holes are attracted by p- and n-sides of the junction, respectively. Due to the high doping level at the n-side of a p-type diode, a lot of crystal imperfections or “recombination centres” are present at that side, which leads to a high probability of recombination for the holes. So only the minority charge carriers, electrons in this case, contribute to the ionisation signal. Due to this process the charge equilibrium between n- and p-sides of the diode is disrupted by the radiation. When connecting both sides externally to each other, a current will be detected at radiation, which, when the diode is in the unbiased mode, is proportional to the number of electron-hole pairs produced, i.e. to the dose.

### III.2. Equipment

#### III.2.1. Diodes

In radiotherapy, most diodes in use for *in vivo* dosimetry are silicon detectors (Figure III.2). p-Type diodes have been described to suffer less sensitivity loss as a function of accumulated dose than their n-type counterparts (§ III.3.4.2) and to be also less dose-rate dependent (§ III.3.5). However some manufacturers have modified their n-type detectors in order to adapt their characteristics for *in vivo* measurements (Table III.1). Complex interplay between the different factors of influence precludes p-type diodes to be systematically superior to their n-type counterparts (AAPM 2005; Jornet et al, 2000; Shi et al, 2003) and as
a consequence the user is invited to carefully investigate the characteristics of the diode
types, p- as well as n-type, on the base of the available information from vendors and current
publications.

Table III.1: Main characteristics of diode systems dedicated to patient monitoring commercially
available in 2003 (Marinello, in press).

<table>
<thead>
<tr>
<th>COMPANY</th>
<th>ELECTROMETERS</th>
<th>DETECTOR NAME</th>
<th>TYPE</th>
<th>RADIATION QUALITY</th>
<th>BUILD-UP CAP MATERIAL</th>
<th>THICKNESS (g/cm$^2$)</th>
<th>Pre-irradiation dose</th>
</tr>
</thead>
<tbody>
<tr>
<td>INVISION and NUCLEAR ASSOCIATES (Presently Cardinal Health Management Services)</td>
<td></td>
<td>Veridose</td>
<td>n or p</td>
<td>1-4 MV</td>
<td>Brass</td>
<td>0.732</td>
<td>20 kGy with high energy electrons</td>
</tr>
<tr>
<td></td>
<td></td>
<td>PDMQC* (5 channels)</td>
<td>n or p</td>
<td>5-11 MV</td>
<td>Brass</td>
<td>1.369</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>30-471</td>
<td>n or p</td>
<td>12-17 MV</td>
<td>Tungsten</td>
<td>2.606</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>30-472</td>
<td>n or p</td>
<td>18-25 MV</td>
<td>Tungsten</td>
<td>3.574</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>30-473</td>
<td>n or p</td>
<td>5-25 MeV elec</td>
<td>none</td>
<td>0.284</td>
<td></td>
</tr>
<tr>
<td>PTW</td>
<td>VIVODOS</td>
<td>T60010L</td>
<td>p</td>
<td>1-5 MV</td>
<td>Titanium</td>
<td>1</td>
<td>Not</td>
</tr>
<tr>
<td></td>
<td></td>
<td>T60010M</td>
<td>p</td>
<td>6-12 MV</td>
<td>Lead</td>
<td>2</td>
<td>Pre-irradiated</td>
</tr>
<tr>
<td></td>
<td></td>
<td>T60010H</td>
<td>p</td>
<td>15-25 MV</td>
<td>Tungsten</td>
<td>3</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>T60010E</td>
<td>p</td>
<td>electrons</td>
<td>PPMA</td>
<td>0.35</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>INVISION Veridose</td>
<td>n or p</td>
<td>organs at risks</td>
<td>PVC +epoxy</td>
<td>0.5</td>
<td>8 kGy delivered with 10 MeV electrons</td>
</tr>
<tr>
<td></td>
<td></td>
<td>PDMQC*</td>
<td>n or p</td>
<td>cobalt-60</td>
<td>Polyst+epoxy</td>
<td>0.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>(5 channels)</td>
<td>p</td>
<td>4-8 MV</td>
<td>Stainless steel</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>p</td>
<td>6-12 MV</td>
<td>Stainless steel</td>
<td>1.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>p</td>
<td>10-20 MV</td>
<td>Stainless steel</td>
<td>2</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>p</td>
<td>≥ 16MV</td>
<td>Tantal</td>
<td>1.7</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>p</td>
<td>electrons</td>
<td>Epox</td>
<td>0.2</td>
<td></td>
</tr>
<tr>
<td>IBA (formerly SCANDITRONIX-WELLHÖFER)</td>
<td>DPD 3 (3 channels) or DPD 510 (10 channels)</td>
<td>EDD-5</td>
<td>p</td>
<td>organs at risks</td>
<td>PVC +epoxy</td>
<td>0.5</td>
<td>8 kGy delivered with 10 MeV electrons</td>
</tr>
<tr>
<td></td>
<td></td>
<td>EDPP-5</td>
<td>p</td>
<td>cobalt-60</td>
<td>Polyst+epoxy</td>
<td>0.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>EDP-10</td>
<td>p</td>
<td>4-8 MV</td>
<td>Stainless steel</td>
<td>1</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>EDP-15</td>
<td>p</td>
<td>6-12 MV</td>
<td>Stainless steel</td>
<td>1.5</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>EDP-20</td>
<td>p</td>
<td>10-20 MV</td>
<td>Stainless steel</td>
<td>2</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>EDP-HL</td>
<td>p</td>
<td>≥ 16MV</td>
<td>Tantal</td>
<td>1.7</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>EDD-2</td>
<td>p</td>
<td>electrons</td>
<td>Epox</td>
<td>0.2</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>DPD-12pc</td>
<td>p</td>
<td>organs at risks</td>
<td>PVC +epoxy</td>
<td>0.5</td>
<td>8 kGy delivered with 10 MeV electrons</td>
</tr>
<tr>
<td>SUN NUCLEAR</td>
<td>IVD or RF-IVD (3 channels)</td>
<td>QED 111300</td>
<td>n or p</td>
<td>70 kV and up</td>
<td>aluminium</td>
<td>0.11</td>
<td>10 kGy delivered with 10 MeV electrons</td>
</tr>
<tr>
<td></td>
<td></td>
<td>QED 111400</td>
<td>n or p</td>
<td>1-4 MV</td>
<td>brass</td>
<td>1.03</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>QED 111500</td>
<td>n or p</td>
<td>6-12 MV</td>
<td>brass</td>
<td>1.85</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>QED 111600</td>
<td>n or p</td>
<td>15-25 MV</td>
<td>acrylic</td>
<td>3.04</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>QED 111200</td>
<td>n or p</td>
<td>electrons</td>
<td>acrylic</td>
<td>0.3</td>
<td></td>
</tr>
</tbody>
</table>

* Can also be connected to a linac quality control device

Most companies offer detectors which are covered with some build-up material of different thickness and nature (Table III.1). In practice it is important to know the characteristics of the build-up cap fixed to the diode prior to measurements, because some caps might not be designed according to the recommendations in chapter II. In order to reduce the physical thickness of the build-up cap, higher density materials (e.g. stainless steel) are used for diodes to be used at high photon energies.
Figure III.2: Example of detector design. The diodes are connected to a central box, which is inside the treatment room and connected to an electrometer situated outside the room.

Some companies perform a pre-irradiation at high dose in order to reduce the variation of the response of the diode with accumulated dose, especially during its initial clinical use (§ III.3).

### III.2.2. Electrometers

The electrometers, to which the *diode dosimeters* are connected, provide a minimum of 2 channels, which allows both an entrance and an exit dose measurement. Other types of electrometers with more than 2 channels, up to a maximum of about 10, are also commercially available. For special techniques, requiring a large number of simultaneous dose measurements, e.g. for TBI, this large number of channels available is very useful.

Computerised electrometer models are also available. Some are equipped with a calibration mode, which allows the signal of the detector to be displayed directly in absorbed dose. Moreover these models often provide the possibility to introduce, for each detector, a correction factor to account for differences between calibration and clinical conditions. This method is convenient only when all conditions remain the same or have only very limited variations.
III.3. Basic characteristics

Several basic characteristics discussed below, mainly the dependence of diode response on accumulated dose (§III.3.4.2), dose-rate (§III.3.5) and temperature (III.3.6), are known to be related to some crystal characteristics, e.g. the doping level. In a continuous effort for further improvements, those characteristics are still being adapted by some companies. The data presented in this work have been taken from the scientific literature as presently available.

III.3.1. Signal stability after irradiation

The stability of the signal which is displayed after irradiation should be checked for time periods corresponding to those encountered in clinical practice. The drift should not exceed 1% in 1 hour when used for conventional treatments. For particular applications such as integration of the dose during total body irradiation performed at low dose-rate, the checks should be extended to a period of several hours.

Practical consequences

Diodes and associated electrometer should have a drift less than 1% in the time range encountered in practice. Equipment falling outside this range should not be accepted.

III.3.2. Intrinsic precision

The intrinsic precision is given by the reproducibility of the signal for at least 10 consecutive irradiations at the same dose. The standard deviation should not exceed 1%.

Practical consequences

For diodes connected to electrometers of good quality the standard deviation should be less than 1%.

III.3.3. Sensitivity: identification of diodes

As relatively large differences in sensitivity may occur between diodes, it is important to identify them. For most modern equipment an identification number is labelled on the diodes, but it is not always the case.

Practical consequences

If the diodes are not labelled by the company they should be carefully identified by the user.
III.3.4. Influence of dose

III.3.4.1. Response variation with dose

The response of a diode has been shown to remain proportional to the absorbed dose, up to doses of more than 10 Gy (Figure III.3).

![Graph showing the response of a Scanditronix p-type EDE diode as a function of dose in the range between 1 Gy and 11 Gy (Van Dam, unpublished results).](image)

Figure III.3: Response of a Scanditronix p-type EDE diode as a function of dose in the range between 1 Gy and 11 Gy (Van Dam, unpublished results).

**Practical consequences**

*No correction for non linearity of the signal is necessary in the dose range encountered for in vivo measurements.*

III.3.4.2. Effect of accumulated dose on detector sensitivity

It is well known that a diode has some “memory” of the dose which it gradually accumulates over its lifetime through clinical use. This accumulated dose results indeed in a progressive decrease in detector sensitivity. This is due to additional lattice defects, produced by the radiation, which act as recombination centers for the minority charge carriers. The loss in sensitivity for a diode as a function of the dose “slowly” accumulated over its lifetime was studied by Larin (1968), Rikner and Grusell (1983) and Van Dam et al (1990). Their findings can be summarized as follows:

– the initial loss in sensitivity of a diode as a function of dose accumulated can be rather steep, especially for new diodes (Figure III.4);
– the phenomenon is more pronounced for the n- than for p-type diodes investigated (Figure III.4);
– the loss of sensitivity is more important for high energy electrons than for photons and, for the same type of particle, the effectiveness increases as a function of energy.

Figure III.4: Sensitivity decrease of a p- and an n-type detector irradiated in a $^{60}$Co beam as a function of pre-irradiation dose with 20 MeV electrons (After Rikner and Grusell, 1983).

For diodes sold pre-irradiated by the company, the decrease of sensitivity as a function of accumulated dose is of course less pronounced. If it is not the case, a pre-irradiation by the user is recommended and may be performed at high dose-rate. For instance p-type diodes have been found to have lost their steep decrease in sensitivity as a function of dose after an irradiation of 20 kGy with 18 MV photons or 5 kGy with 20 MeV electrons (Fig.III.4).

**Practical consequences**

*The sensitivity of diodes should be checked periodically, especially when they are new.*

*It is recommended to pre-irradiate the diodes at high dose (several kGy) before use, or to buy pre-irradiated diodes.*
III.3.5. Influence of dose-rate

During the irradiation electron-hole pairs are produced at a rate which increases as a function of dose-rate (dose per unit time). If this dose-rate is high, which is the case for pulsed radiation produced by linacs (because of the very high dose-rates present within the pulse), a phenomenon of “pile up” occurs: i.e. the ions are produced at such a high rate that the recombination cannot “keep pace” and more charge carriers escape recombination than at lower dose-rates. As a consequence diode sensitivity decreases when decreasing dose per radiation pulse and this phenomenon has been observed to be more important for n- than for p-type diodes (Grusell and Rikner, 1984). New p-type detectors, with a higher doping level, have only a very limited dose-rate dependence (Grusell and Rikner, 1993; Ding et al, 1995; Shi et al, 2003), while old detectors, after accumulating dose through clinical use, may show significant response increase as a function of dose-rate (Van Dam et al, 1990).

As, in clinical practice, it seems hardly realistic to take the exact dose-rate into account each time, a simplified approach consists of calibrating the diodes in the dose-rate range in which they will be used on the patient:

– diodes which will be used for entrance dose measurements could be calibrated e.g. at isocentre and for a reference collimator opening; the same calibration factor may then be applied to isocentric techniques or other collimator openings;
– diodes which will be used for exit dose measurements could be calibrated in exit position, for an average patient thickness, with the entrance surface of the beam at isocentre and again for a reference collimator opening;
– diodes to be used for e.g. TBI at low dose-rate, which dose-rate is achieved by increasing the SSD, should be calibrated in these conditions; due to the marked dose-rate decrease compared to the usual value (a factor of 50 is not uncommon), the application of the calibration factor determined in the usual conditions could introduce an error of some 10%, which is quite unacceptable.

Moreover there are a few situations where the dose-rate effect merits special attention such as:

– the use of a wedge filter decreases the dose-rate and consequently change the diode response. For some diodes it may cause a loss of accuracy of up to 2% for the largest wedge angles;
– particular care is also required for users of linacs which offer the choice between different monitor rates by variation of the pulse intensity (not of the pulse repetition frequency);
– finally the dose-rate effect is especially critical in linac electron beams, for which field flatness is achieved by magnetically scanning a pencil beam, of very high intensity, over the field. Indeed, for such beams, when the diode is hit by the pencil beam, very high instantaneous dose-rates are reached at its level.
III.3.6. Influence of temperature

Most authors (Grusell and Rikner, 1986; Van Dam et al, 1990; Heukelom et al, 1991) report an increase of diode response with temperature. The basic explanation for an increase in response is again at the level of the minority charge carriers: when increasing temperature their energy, and thus their probability for “escaping” recombination, also increases. In the range between room and skin temperature (about 32°C) a linear increase of diode sensitivity as a function of temperature has been observed (Fig.III.5). It should be noted that before accumulating dose, diodes have a rather moderate temperature dependence, with a sensitivity variation with temperature (SVWT) below 0.2 % per °C. However, when they start to accumulate dose through clinical use, an initial sharp rise in SVWT is observed, which might level off at a plateau as high as 0.4 % per °C. Similarly to the decrease in response (§III.3.4.2), the increase in SVWT as a function of accumulated dose will be reduced, or even eliminated when the diodes are sold pre-irradiated by the company.

Figure III.5: Responses of six different p-type Scanditronix diodes as a function of temperature between 20°C and 32°C. At the time of the measurements, diodes 1 to 5 and diode 6 had accumulated doses of 4 kGy and 300 Gy respectively. In the temperature range explored a linear increase of diode response as a function of temperature was observed (Van Dam et al, 1990).
The temperature dependence of a diode is probably the characteristic which is the most difficult one to master in clinical conditions. Different methods have been proposed in order to take this effect into account: from performing the calibration in a phantom filled with water at skin temperature to eliminating electronically the response variation as a function of temperature (Klevenhagen, 1978). One of the main difficulties is, in most clinical conditions, the impossibility to know the temperature of the diode during patient irradiation. Usually after patient set-up the diode is taped on the skin and, immediately after, irradiation is switched on. So the time between taping and first cGy is only 10-20 s. On the other hand it has been shown (Grussel and Rikner, 1986) that it takes several minutes for a diode, after being taped on the patient, to reach thermal equilibrium with the skin, which is at about 32°C (Figure III.6). This means that in standard practice diode temperature during irradiation, which is of the order of a minute, is still increasing. This is why the approach of most users is just not to correct for temperature variation between room (calibration conditions) and skin. Moreover, when the diodes are taped on a mask, the temperature is probably not far from room temperature. In order to keep the loss of accuracy to not more than a percent when the detector is used directly on the skin, it is advisable to determine the SVWT of a new diode by the use of a thermostatically controlled water tank and to follow it over time (e.g. check every 6 months). Once the SVWT exceeds 0.4% per °C, the diode should be eliminated. Even with this cut-off value, the temperature effect of diodes remains one of the main causes for a decrease of accuracy, which however can be accepted in most clinical applications.

Figure III.6: Variation of diode temperature as a function of time after taping on the patient (After Grusell and Rikner, 1986).
The problem of temperature dependence of diode response has however a very straightforward solution in the (rare) cases where the irradiation of the patient takes place at low dose-rate. A typical example is total body irradiation at low dose-rate (less than 10 cGy per minute), which is sometimes applied in single fraction regimen. Moreover the taping of the detectors at the different loci investigated takes time, so that the detectors have plenty of time to adapt their temperature to that of the skin before start of the irradiation.

### Practical consequences

*Sensitivity variation with temperature (SVWT) should be determined for each new diode and be checked periodically. As long as it remains smaller than 0.4% per °C, no temperature correction is needed for in vivo dosimetry.*

### III.3.7. Influence of energy

Due to the lack of tissue equivalence of a diode, its response varies as a function of energy for photon as well as for electron beams.

#### III.3.7.1. Photon energy

A diode has a much higher atomic number Z than that of soft tissue: e.g. the Z of a silicon diode is about 14, while that of soft tissue is about 7. Due to this difference, the photoelectric effect is more important in the diode and its response increases with decreasing energy below cobalt 60. This was indeed confirmed experimentally (Heukelom et al, 1991) and is of particular concern for conventional X-rays (Fig.III.7). When high-density build up caps are used, this effect is even reinforced.

The energy dependence of a diode being very hard to calculate the only way is to measure it (by comparison with an ionization chamber) for every energy for which the diode will be used. Thanks to the relatively small dose response variation as a function of energy, spectral variations produced by changes in collimator opening, SSD, etc. are negligible.

#### III.3.7.2. Electron energy

The dependence of the mass stopping power ratio on electron energy for water to air being different from that for water to e.g. silicon (Rikner, 1985), diode response is expected to vary as a function of electron energy. Moreover it is mandatory to avoid high density build-up caps in the case of electron beams because of the major spectral and scattering variation introduced by such caps at the diode level.
III.3.8. Directional effect

Constructional anisotropies in the diode result in sensitivity variations as a function of the angle between the central beam axis and the symmetry axis of the diode and of the build-up cap (Figure III.8). The more “oblique” the radiation (i.e. the larger the angle between beam axis and diode symmetry axis) the more important this variation, which may amount to more than 5% for angles of 60° (Shi et al, 2003).

Practical consequences

The response of a diode and associated build-up cap should be measured as a function of energy in the photon or electron energy range in which it is used.

When measuring outside the field or under shielding blocks, particular attention should be paid to energy dependence of diode response, because of the large spectral differences with in-field positions.

III.3.8. Directional effect

Constructional anisotropies in the diode result in sensitivity variations as a function of the angle between the central beam axis and the symmetry axis of the diode and of the build-up cap (Figure III.8). The more “oblique” the radiation (i.e. the larger the angle between beam axis and diode symmetry axis) the more important this variation, which may amount to more than 5% for angles of 60° (Shi et al, 2003).

Practical consequences

Directional dependence of diode response may be of importance when large angles may have to be dealt with: e.g. in the tangential irradiation of the breast or the thoracic wall. Also when performing off-axis measurements the angular dependence of the diode response could be of importance. It is then the task of every user of diodes to study the directional effect for his own detectors and to evaluate its importance in the clinical conditions explored.
III.4. Clinical use

The application of the basic characteristics of diodes for in vivo dosimetry is focused on the relative constancy of their response as a function of time. In that respect they show a striking similarity with ionization chambers, although their possible decrease in sensitivity as a function of dose accumulated is obviously a complicating factor. Nevertheless, it makes sense to attribute to a diode an individual calibration factor, although it will be necessary to monitor this factor over time more closely than for an ionization chamber. Moreover, in the case of diodes, the influence of some factors such as dose-rate, temperature, etc. constitutes a possible supplementary complication. Despite these drawbacks diodes, when used in skilful hands, have proven very useful in in vivo dosimetry, making an important contribution to quality assurance.

Figure III.8: Relative response of 3 different diodes, used at different photon energies (without additional build-up cap), as a function of the angle $g$ between the symmetry axis of the diode and the beam axis. For the 3 diodes investigated the sensitivity decreases as a function of the angle (After Van Dam et al, 1994).
III.4.1. Calibration factors of a diode

It is strongly recommended to perform a separate calibration for each beam in which a diode is used for in vivo dosimetry. As a matter of fact it turns out to be very convenient in practice, when diodes are used extensively in different beams in a department, to assign to each treatment unit an electrometer and a number of diodes. This of course limits the number of beams in which a diode must be calibrated.

If a diode is used for entrance and exit dose measurements, it has to be ascribed a corresponding entrance as well as exit dose calibration factors.

III.4.1.1. Entrance dose calibration factor

The entrance dose calibration factor \( F_{\text{entrance}} \) is defined as the factor, valid in reference conditions, with which the signal \( R_{sc,\text{entrance}} \) of the diode, positioned on the skin of the patient at the entrance surface with the build-up cap, must be multiplied to yield the entrance dose \( D_{\text{entrance}} \) (§ II.1). The entrance dose calibration factor is given by:

\[
F_{\text{entrance}} = \frac{D_{\text{entrance}}}{R_{sc,\text{entrance}}} \quad [5]
\]

For determination of \( F_{\text{entrance}} \), the diode is put on the surface of a flat phantom, at the entrance side of the beam, and its response is compared with that of a calibrated ionisation chamber, positioned at depth \( d_{\text{max}} \) (Figure III.9).

If the diode is used for a particular kind of treatment, e.g. mantle fields, it is recommended to take an average set-up (collimator opening, SSD, etc) for the treatment involved as reference conditions. But if the application field is much wider, e.g. entrance dose measurement for every new patient, it is obvious that the reference conditions (e.g. collimator opening 10 cm x 10 cm, phantom surface at isocenter) are representative for an application range with a broader variability in the geometrical parameters. It is recommended to check for the main treatment techniques applied, the validity of this one single calibration factor (e.g. check for a representative head-and-neck field, for a mantle field). For some special techniques such as total body irradiation, it is recommended to make a separate calibration in conditions close to those applied during this kind of application.

III.4.1.2. Exit dose calibration factor

Similarly, an exit dose calibration factor \( F_{\text{exit}} \) can be determined for a diode dosimeter by putting it on the exit surface of the beam and by comparing its response with that of an ionization chamber positioned in the phantom at \( d_{\text{max}} \) from this surface (Fig.III.9).
Figure III.9: Determination of the entrance and exit dose calibration factors of a detector (diode or TL dosimeter), $F_{\text{cal, en}}$ and $F_{\text{cal, ex}}$, respectively. The calibrated ionisation chamber is put at depth $d_{\text{max}}$ for the entrance dose calibration factor (A) or at a distance $d_{\text{max}}$ from the exit side of the field for the exit dose calibration (B) and the detector is positioned at the entrance or exit surface, respectively. In order to avoid the shadow of the entrance detector on the ionization chamber, it is recommended to shift slightly the detectors with respect to each other.

The exit dose calibration factor $F_{\text{exit}}$ is given by:

$$F_{\text{exit}} = \frac{D_{\text{exit}}}{R_{\text{sc, exit}}}$$  \[6\]

in which the symbols, except for “exit”, have the same meaning as for the entrance dose calibration factor.

With regard to the reference conditions adopted for exit dose calibration, patient thickness constitutes an additional parameter. The calibration phantom thickness should preferably equal the average thicknesses encountered in clinical conditions. A check of the single exit calibration factor is also recommended for 2 or 3 typical thicknesses.

An important parameter is, for diodes which are used for both entrance and exit dose measurements, the ratio $F$ between exit and entrance dose calibration factor:

$$F = \frac{F_{\text{exit}}}{F_{\text{entrance}}}$$  \[7\]
Due to the difference in relative positions of ionisation chamber and diode dosimeter for entrance and exit dose determination, and also because of the lower dose-rate at the exit side of the beam (§ III.3.5), F is expected to be larger than unity. However, for some types of diodes, surrounded by a hemispherical build-up cap and mounted on a thin plate (Figure III.2) F values of up to 1.12 have been observed. In no way can these high values be explained by differences in position between ionisation chamber and diode or by dose-rate effects. They are related to the fact that diodes of this type have to be reversed for exit dose measurements. Complementary measurements have shown that the decrease in response observed when just reversing this orientation, without changing the distance, is the main origin of the relatively high $F_{\text{exit}}$ values observed. This is due to anisotropies in the construction of the diodes (AAPM, 2005). For such diodes it is then not allowed just to use the entrance dose calibration factor for exit dose measurements.

Because the dose accumulated over the life time of a diode decreases its response (§ III.3.4.2), the calibration factors increase while the detector accumulates dose during clinical use. However, it has been shown (Leunens, 1990-b) that the relative increases of entrance and exit calibration factors are equal, so that their ratio F does not vary over time. It is then adequate to follow the time evolution of only one of the two calibration factors. The frequency of the checks depends on that of the use of the detector as well as on the accuracy required for the measurements.

### III.4.2. Use of a diode in clinical conditions

Every parameter which influences diode response (§ III.3) intrinsically may be the reason why, when the clinical conditions are different from the reference conditions, the calibration factor alone is not more adequate to convert the detector signal into absorbed dose. A number of correction factors are then necessary (Table III.2). They are introduced as cumulative factors with which the calibration factors have to be multiplied in order to get the correct dose: if the response of the detector is lower or higher in clinical than in reference conditions, these factors are respectively larger or smaller than unity. Some of the values recommended are strongly related to the methodology adopted which, for reminder, is indicated between brackets in Table III.2. When the value of the correction factor is equal to unity, it may be either because basically the corresponding parameter has no influence on detector response or because the error is estimated not to exceed 1 %.

Use of diodes for in vivo dosimetry in IMRT is possible (Higgins et al, 2003) but is not treated in present work, because more experience is required before recommendations in this field can be issued.
Table III.2: Values of the different correction factors to be applied in clinical conditions for diodes and radiothermoluminescent dosimeters.

<table>
<thead>
<tr>
<th>Correction factors (Parameter)</th>
<th>Value for</th>
<th>Value for</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>diodes (conditions)</td>
<td>TLD (conditions)</td>
</tr>
<tr>
<td><strong>C_D</strong> (dose)</td>
<td>1</td>
<td>&lt; 1</td>
</tr>
<tr>
<td></td>
<td>(in linear region of dose/response curve)</td>
<td>(in supralinear region)</td>
</tr>
<tr>
<td><strong>C_{DR}</strong> (dose-rate)</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>(calibration per dose-rate range: entrance, exit, TBI)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>&gt; 1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>(for wedge)</td>
<td></td>
</tr>
<tr>
<td><strong>C_T</strong> (temperature)</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>(in usual conditions provided SVWT &lt; 0.4% per °C)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>&lt; 1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>(TBI)</td>
<td></td>
</tr>
<tr>
<td><strong>C_E</strong> (energy)</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>(separate calibration per beam)</td>
<td>(calibration detectors irradiated with same radiation quality as patient detectors)</td>
</tr>
<tr>
<td><strong>C_{DI}</strong> (direction)</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>(close to field centre)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>≠ 1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>(tangential field or non-central position)</td>
<td></td>
</tr>
</tbody>
</table>
IV. RADIOTHERMOLUMINESCENT DOSIMETERS

Thermoluminescence dosimetry (TLD) has been developed considerably over the past ten years, the commercial availability of reliable detector materials and the commercialization of automatic readout systems being a decisive factor. For in vivo measurements TL detectors have the advantage of being highly sensitive under a very small volume and not to have to be connected to an electrometer with an unwieldy cable. Their major disadvantage which is the time required for readout can be considerably decreased by a good choice of the equipment and a good methodology.

IV. 1. Principle of the method

Thermoluminescence dosimetry is based upon the ability of imperfect crystals to absorb and store the energy of ionizing radiation, which upon heating is re-emitted in the form of electromagnetic radiation, mainly in the visible wavelength. The light emitted is then detected by a photomultiplier (P.M.) and correlated to the absorbed dose received by the TL material (Mc Kinlay, 1981; Mc Keever, 1985).

One of the possible mechanisms is presented in figure IV.1. Energy states in a crystal being represented with energy increasing upwards along the ordinate, free electrons and holes are produced under the irradiation effect. Both of them are free to travel through the solid in the conduction band for a short time. They may be ultimately either trapped at defects or fall back into the valence band and recombine either radiatively (fluorescence) or non radiatively with holes, or be captured at luminescent centers with the emission of light. The electrons may stay in the traps for prolonged periods (up to months), which confers to the thermoluminescent method the advantage of storage of information.
The information can then be collected by heating the crystal to a temperature depending upon its nature. The calorific energy is used by the electrons to escape from the trap again into the conduction band, where they are free to travel and have three possible fates as before: either be retrapped at defects or fall into the valence band and recombine radiatively or not with holes, or recombine radiatively at a hole-activated luminescence center. It is the light emitted by this last process which is called thermoluminescence (TL).

Heating and light collection are performed in a readout system called the reader. The TL signal as a function of temperature (or of time if this parameter is correlated with temperature) is of a complex nature and is called a glow curve. It consists of different TL peaks, each peak corresponding to a different energy state in the crystal (Figure IV.2). They are either unstable, decaying more or less quickly with time according to the TL material considered, or stable. A TL dosimeter always contains both unstable and stable peaks, the latter being the one(s) used in dosimetry. They are called dosimetric peaks.

![Glow curve of lithium borate doped with manganese composed of two peaks.](image)

Figure IV.2: Glow curve of lithium borate doped with manganese composed of two peaks. The low temperature peak decays rapidly at room temperature whereas the dosimetric peak remains stable. The unstable peak was read out 3.4 min (A), 8.6 min (B), 27.3 min (C), 69.2 min (D) and 900 min (E) after irradiation (After Schulman et al, 1967).

After readout, the TL material is either entirely in its original state, and in this case it is just ready for re-use, or it requires a special heating treatment called annealing in order to restore it to its original state (§ IV.3.4.2).
IV.2. Equipment

IV.2.1. TL dosimeters

Most commonly used TL detectors are obtained by doping phosphors such as lithium fluoride (LiF), lithium borate (Li₂B₄O₇), calcium sulphate (CaSO₄) and calcium fluoride (CaF₂) with impurities called activators: e.g. LiF:Mg-Ti is lithium fluoride doped with magnesium and titanium, Li₂B₄O₇:Cu is lithium borate doped with copper, etc. All TL materials are available either in the form of powder or of solid dosimeters (Figure IV.3). The solid dosimeters may be made entirely of phosphors as single crystals or polycrystalline extrusions (extruded rods, sintered pellets or chips), or as homogeneous composites of the phosphor powder and some binding material. It should be noted that the characteristics of the pure phosphor dosimeters may be considerably different from those of the composites.

Figure IV.3: Example of TL materials. The coin (2.1 cm in diameter) gives an idea of their dimensions.

Among available TL materials those which can be considered equivalent to soft tissues or to bones in the energy range encountered in radiotherapy are listed in Table IV.1.
Table IV.1: Different TL materials equivalent to soft tissues, lungs or bones.

<table>
<thead>
<tr>
<th>Soft tissue or lung</th>
<th>Bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>LiF (Mg, Ti)</td>
<td>CaSO₄ : Mn</td>
</tr>
<tr>
<td>LiF (Mg, Ti, Na)</td>
<td>CaSO₄ : Dy</td>
</tr>
<tr>
<td>Li₂B₄O₇ : Mn</td>
<td>CaF₂ : Mn</td>
</tr>
<tr>
<td>Li₂B₄O₇ : Cu</td>
<td>CaF₂ : Dy</td>
</tr>
</tbody>
</table>

**IV.2.2. Readers**

Their principle is shown in figure IV.4. Most often the irradiated dosimeter is placed within a metallic support (or tray or planchette) situated within a readout chamber. The support is heated by different means at two different temperatures: the preheating temperature used to clear unstable peaks and the readout temperature used to collect the information from dosimetric peaks (Figure IV.2). Generally the temperature of the support is measured by a thermocouple in close thermal contact with it. The readout chamber should be continuously flushed with nitrogen gas in order to reduce spurious phenomena (McKeever, 1985), and thus decrease the background. The light emitted by the hot TL material passes through one or more optical filters before being detected by a photomultiplier tube (PM). The output current from the PM is proportional to the light emission, and therefore, after integration, to the absorbed dose previously received by the TL dosimeter.

Figure IV.4: Schematic diagram of a classical TLD reader.
Different heating systems are encountered in TLD readers commercially available (Table IV.2):

– the metallic support containing TL material may be heated by an electric current, or through contact with ovens, or a hot finger moved by a lift mechanism. Heating kinetics are either linear or isothermal as a function of time. When it is linear, the TL material is progressively heated to preheating and readout temperatures. When it is isothermal, the TL material is quasi-instantaneously heated by isothermal ovens to both of these temperatures and generally readouts take less than 10 seconds. In any case close contacts between TL dosimeter, support and heating system are necessary to obtain a good reproducibility;

– some non-contact procedures, such as heating by hot nitrogen gas or air, optical infrared heating using an intense light pulse from an halogen lamp or heating by a laser beam (Gasiot et al, 1982; Kearfott and Grupen-Shemansky, 1990), can also be used. In these cases the heating kinetics are particular and often difficult to establish.

Table IV.2: Some TL readers commercially available in 2005.

<table>
<thead>
<tr>
<th>READER MAKERS</th>
<th>MODEL</th>
<th>MANUAL or AUTOMATIC</th>
<th>HEATING PROCEDURE</th>
<th>MAXIMUM TEMPERATURE (°C)</th>
<th>LIGHT DETECTION</th>
<th>NATURE</th>
<th>FORM</th>
</tr>
</thead>
<tbody>
<tr>
<td>THERMO-ELECTRON (USA)</td>
<td>HARSHAW 3500</td>
<td>Manual</td>
<td>Planchet</td>
<td>400°C</td>
<td>Hamamatsu bialkali PMT, fixed optical filter</td>
<td>LiF, CaSO₄</td>
<td>Powder and solid detectors</td>
</tr>
<tr>
<td>THERMO-ELECTRON (USA)</td>
<td>HARSHAW 4500</td>
<td>Manual</td>
<td>Planchet and hot nitrogen gas</td>
<td>400°C (Planchet) or 600°C (gas)</td>
<td>Hamamatsu bialkali PMT, fixed optical filter</td>
<td>LiF, CaSO₄</td>
<td>Powder, solid detectors, kapton cards</td>
</tr>
<tr>
<td>THERMO-ELECTRON (USA)</td>
<td>HARSHAW 5500</td>
<td>Automatic 50 dosi/25 min</td>
<td>Hot nitrogen gas</td>
<td>400°C</td>
<td>Hamamatsu bialkali PMT, fixed optical filter</td>
<td>LiF, CaSO₄</td>
<td>Solid dosimeters</td>
</tr>
<tr>
<td>THERMO-ELECTRON (USA)</td>
<td>N.E. TECHNOLOGY RIALTO</td>
<td>Automatic 30 dosi/10 min</td>
<td>Planchet</td>
<td>400°C</td>
<td>Hamamatsu bialkali PMT, fixed optical filter</td>
<td>LiF, CaSO₄</td>
<td>Powder, solid detectors, extremity tapes</td>
</tr>
<tr>
<td>FIMEL (France)</td>
<td>LTM</td>
<td>Manual</td>
<td>Planchet</td>
<td>500°C</td>
<td>Hamamatsu bialkali PMT + interchangeable optical filters</td>
<td>All TL material</td>
<td>Powder and solid detectors</td>
</tr>
<tr>
<td>FIMEL (France)</td>
<td>PCL 3</td>
<td>Automatic 45 dosi/15 min</td>
<td>2 isothermal fingers</td>
<td>600°C</td>
<td>Hamamatsu bialkali PMT + interchangeable optical filters</td>
<td>All TL material</td>
<td>Powder and solid detectors</td>
</tr>
</tbody>
</table>

Readers which are designed for the readout of a great number of dosimeters in a short time generally have isothermal heating kinetics (Marinello et al, 1992) or heat the TL dosimeters in hot gas.
The TL light detection is variable from one reader to another because it depends on
the composition of the PM photocathode and on the spectral transmission of its tube window.
In most readers photocathodes are bialkalis with a peak sensitivity around 400 nm in good
agreement with the blue emission of LiF or Li$_2$B$_4$O$_7$:Cu, but not necessarily adapted for the
yellow-red light of Li$_2$B$_4$O$_7$:Mn. Adequate filters placed in front of the PM window allow the
reader to be adapted to the wavelength of any TL dosimeter to be used. A good reader should
have a PM with a large spectral transmission and should allow a quick interchange of the
associated filters in order to be adaptable to different TL materials.

Depending upon the type of the readout system used, the signal proportional to
the light emission is either amplified and fed to an integrator (d.c operation regimen), or
converted into pulses and fed to a scaler (pulse counting regimen). Irrespective of the
regimen the voltage of the PM tube has to be correctly stabilised in order to get a good
reproducibility of measurements.

Results which have to be correlated to absorbed dose are either read out and stored
by the operator, or safeguarded on the hard disk of a computer and then used and printed
under the form wished by the user. In such case the different calibration and correction
factors can be taken into account automatically. In many readers glow curves are
also displayed during dose measurements so as to provide a maximum amount of
information.

IV.3. Basic characteristics

IV.3.1. Signal stability after irradiation

An important consideration in the choice of a TL dosimeter is the stability of the
signal. In particular it is necessary to assess whether the charges trapped during the
irradiation have not been lost before the readout by unwanted exposure to heat (thermal
fading), light (optical fading) or any other factor (anomalous fading). This is expressed by a
decrease of the TLD response depending on the delay separating irradiation and readout.

An appropriate preheating allows the elimination of that part of the signal (low
temperature peaks) which presents an important thermal fading, and then reduces
considerably thermal fading for most TL materials (Table IV.3). Except for older styled
readers the preheating is part of the readout cycle. Thus Li$_2$B$_4$O$_7$:Mn, which shows a
significant fading without preheating (Figure IV.2), has a thermal fading of the order of only
2.5 % per month when it is correctly preheated. Likewise LiF shows a fading of the order of
5 to 10% per year depending upon its manufacture and its previous annealing (Portal, 1981;
Mc Keever, 1985).

Optical fading can be avoided by manipulating the dosimeters in a room illuminated
with incandescent light and wrapping them in opaque containers or envelopes, when used for
in vivo dosimetry in treatment rooms illuminated with fluorescent light.
Table IV.3  Thermal fading of different TL materials after correct preheating (After Visocekas et al, 1985 for Li$_2$B$_4$O$_7$:Cu and Mc Keever, 1985 for the others)

<table>
<thead>
<tr>
<th>TL Material</th>
<th>Thermal fading</th>
</tr>
</thead>
<tbody>
<tr>
<td>LiF</td>
<td>5 to 10 % per year</td>
</tr>
<tr>
<td>Li$_2$B$_4$O$_7$: Min</td>
<td>2.5 % per month</td>
</tr>
<tr>
<td>Li$_2$B$_4$O$_7$: Cu (C.E.N.-FAR)</td>
<td>4 % per month</td>
</tr>
<tr>
<td>CaSO$_4$: Dy</td>
<td>1 to 5 % per month</td>
</tr>
<tr>
<td>CaSO$_4$: Mn</td>
<td>10 % per month</td>
</tr>
<tr>
<td>CaF$_2$: Dy</td>
<td>25 % per month</td>
</tr>
<tr>
<td>CaF$_2$: Mn</td>
<td>7 % per day</td>
</tr>
</tbody>
</table>

Anomalous fading is much more difficult to detect than either thermal or optical fading because it generally occurs much more slowly. Possibly because of this it has not yet been demonstrated to be a problem for in vivo dosimetry.

**Practical consequences**

Thermal fading should be evaluated on each individual reader with the TL material which is intended to be used. It should be approximately 1% per month, or less, for the different preparations of LiF when readout and annealing conditions are reached. For Li$_2$B$_4$O$_7$ it varies from 0.5 to 1% per week depending upon the doping. When a long delay separates irradiation from readout, a fading correction may be necessary.

IV.3.2. Intrinsic precision

Intrinsic precision is the reproducibility of a given TL material associated with a given readout system. It is very dependent on the quality of the TL material used, reader characteristics, the way in which the preheating and heating cycle have been defined, the purity of the nitrogen gas circulating in the readout chamber, etc. It can be evaluated by randomly taking 10 samples of TL powder or dosimeters out of the same batch and by irradiating them to the same dose. After readout, and annealing procedure when necessary (§ IV.3.4.2), the operation is repeated several times.

When readout parameters have been optimised, a standard deviation of ± 2% or less, can routinely be obtained with either manual or automatic readers of good quality associated with reliable TL materials (Feist, 1988; Kirby et al, 1992; Marinello et al, 1992; Derreumaux et al, 1995).

**Practical consequences**

TL materials which show a standard deviation higher than ± 2 % are either of poor quality or are not correctly handled and read out. They should not be used for in vivo measurements or the handling procedure should be improved.
IV.3.3. Sensitivity: identification of dosimeters

IV.3.3.1. Solid dosimeters

Some variations in sensitivity within a batch of TL dosimeters is unavoidable. Several methods can be used to limit the effect of these variations when the dosimeters are in common use. The best method consists of irradiating all the dosimeters in the same geometrical conditions, to read them out and to attribute to each of them a sensitivity factor $S_i$ equal to:

$$S_i = \frac{R_i}{\bar{R}} \quad [8]$$

where $R_i$ is the TL readout from dosimeter number $i$ and $\bar{R}$ the mean of all values of $R_i$. This sensitivity factor expresses the response variation of each individual dosimeter around the mean. Although this mean may vary from irradiation to irradiation, $S_i$ should remain constant because all dosimeters are subject to the same variations. Sensitivity factors should be checked periodically to take into account a possible loss of material occurring when TL dosimeters are not handled carefully.

Another method giving a similar accuracy consists of dividing the TL dosimeters into sensitivity groups without identifying them individually (e.g. groups of dosimeters with a response variation less than $\pm 1$ or $\pm 2\%$ from the group mean) and to increase the number of dosimeters used for each point of measurement. When an automatic reader is available, such a method is very suitable because readouts take a very short time. As the distribution of sensitivities within a group can vary with time for the same reasons as above, it should be checked at a frequency which depends upon the accuracy required for measurements.

### Practical consequences

**For solid dosimeters: differences in sensitivity should be taken into account either by identifying them and determining their individual sensitivity factor, or by grouping them in batches of similar sensitivity.**

IV.3.3.2. Powder

If optimum accuracy is to be obtained when TL powders are used, the quantity of powder used and the readout conditions must be accurately defined and corrections made when necessary. The response variations with mass of TL material should be established for the readout conditions used in practice because they depend upon the heating kinetics (§ IV.2.2). For most TL materials the signal is proportional to the mass when they are read out with linear heating kinetics: either a linear correction should be made with samples of different weight, or samples of equal weight should be used.

Some TL material such as Li$_2$B$_4$O$_7$:Cu have a response which can be considered as independent of mass, in a certain range of mass, when they are read out with automatic
readers using isothermal heating kinetics (Marinello et al, 1992). In this case it is not necessary to weigh the powder, a simple volumetric measurement being enough (Figure IV.5). Similar observations were made with LiF: Mg,Ti,Na (Derreumaux et al, 1995).

**Practical consequences**

*For powder: response variations with mass of TL material should be established for the readout conditions used in practice and taken into account when necessary.*

![Image of volumetric measurement](image_url)

**Figure IV.5: Volumetric measurement: it consists of completely filling a calibrated hole.**

**IV.3.4. Influence of dose**

**IV.3.4.1. Response variation with dose**

Irrespective of the TL material used, the TL emission per unit of absorbed dose is illustrated by the typical curve shown in Figure IV.6. At relatively low values of absorbed dose the response of the TL material is linear, the origin of the straight line being the background TL signal, $B$, of an unirradiated phosphor. Above a dose $D_l$ limiting the linear region, the response becomes supralinear, saturating at dose $D_s$ and then falling off. This is due to additional lattice defects produced by the irradiation in TL crystals, which results in an increase of detector response as these defects may act as electron traps, and so take part in the TL process. An indication of the dose range corresponding to the linear zone and saturation dose of TL phosphors commonly used in radiotherapy is given in table IV.4. The
occurrence of non linearity in the dose response curve of a TL detector does not preclude its use in TLD, but it requires careful application of correction factors. A good example is given by LiF preparations which are linear only from $10^{-5}$ to 1 Gy (Mc Keever, 1985) although widely used in practice. Nevertheless it is not recommended to use TL dosimeters in the sublinear region approaching saturation.

Table IV.4: Response variation with dose of different TL materials commonly used for in vivo dosimetry. Doses corresponding to the end of the linear region and saturation are presented in column 2 and 3 respectively (After Visocekas et al, 1985 for Li$_2$B$_4$O$_7$:Cu and Mc Keever, 1985 for the others)

<table>
<thead>
<tr>
<th>TL Material</th>
<th>Linear zone</th>
<th>Saturation</th>
</tr>
</thead>
<tbody>
<tr>
<td>LiF</td>
<td>$5 \times 10^{-5}$ to 1</td>
<td>$10^3$</td>
</tr>
<tr>
<td>Li$_2$B$_4$O$_7$: Mn</td>
<td>$10^{-4}$ to 3</td>
<td>$3 \times 10^4$</td>
</tr>
<tr>
<td>Li$_2$B$_4$O$_7$: Cu (C.E.N. - FAR)</td>
<td>$5 \times 10^{-4}$ to 120</td>
<td>$10^3$</td>
</tr>
<tr>
<td>CaSO$_4$: Dy</td>
<td>$10^{-6}$ to 30</td>
<td>$10^3$</td>
</tr>
<tr>
<td>CaSO$_4$: Mn</td>
<td>$10^{-7}$ to 30</td>
<td>$10^2$</td>
</tr>
<tr>
<td>CaF$_2$: Dy</td>
<td>$10^{-5}$ to 10</td>
<td>$10^4$</td>
</tr>
<tr>
<td>CaF$_2$: Mn</td>
<td>$10^{-5}$ to 10</td>
<td>$10^3$</td>
</tr>
</tbody>
</table>

Figure IV.6: Variation of thermoluminescence as a function of absorbed dose

It should also be noted that supralinearity and saturation dose can both be affected by bad heating conditions, by previous exposures to irradiation and by thermal treatments and by the intrinsic response of the reader. (§ IV.3.4.2).
Some TL materials such as LiF are critically dependent upon the thermal treatments, called annealing, performed before their first irradiation and after use (McKeever, 1985). When these treatments are not performed the sensitivity and background of TL dosimeters are considerably altered. Moreover, dosimetric properties do not remain constant. This is due to clustering reactions among defects which can be induced during thermoluminescence production, thereby giving rise to a strong dependence of supralinearity and sensitization properties on heating rates.

Before the first use, the heat treatment for the Li$_2$B$_4$O$_7$, doped with either copper or manganese, consists of an annealing from 15 minutes to 1 hour at 300°C. No further annealing is necessary before re-use.

Annealing is much more complicated for some preparations of LiF and can alter the dosimetric properties of the TL material when not carried out correctly. It depends upon the form in which LiF has been manufactured. The reader is referred to the literature (Portal, 1981; Driscoll et al., 1986) and to indications provided by the manufacturers to adapt the annealing procedures to the LiF preparations used. For instance, Harshaw LiF:Mg,Ti chips or rods should be annealed from 1 to 2 hour at 400°C followed by 24 hours at 80°C after each use.

Some manufacturers recommend no annealing after re-use for the LiF detectors. In our experience correct results can be obtained provided the annealing be replaced by an adequate preheating of the irradiated material just before the readout cycle (Marinello et al., 1992).

**Practical consequences**

In practice it is recommended to use TL dosimeters in the region where their response is proportional to the dose received (linear region). When it is not the case, a correction should be applied to the signal from a curve established with the TL material as well as the reader used (and not from a published curve because the readout parameters may have an influence on its shape). This curve should be checked periodically. It can be introduced in the computer associated with some automatic readout systems, thus allowing an automatic correction of the readouts.

TL dosimeters should not be used in the sublinear region approaching saturation.

---

**IV.3.4.2. Effect of accumulated dose on dosimeter sensitivity**

Some TL materials such as LiF are critically dependent upon the thermal treatments, called annealing, performed before their first irradiation and after use (McKeever, 1985). When these treatments are not performed the sensitivity and background of TL dosimeters are considerably altered. Moreover, dosimetric properties do not remain constant. This is due to clustering reactions among defects which can be induced during thermoluminescence production, thereby giving rise to a strong dependence of supralinearity and sensitization properties on heating rates.

Before the first use, the heat treatment for the Li$_2$B$_4$O$_7$, doped with either copper or manganese, consists of an annealing from 15 minutes to 1 hour at 300°C. No further annealing is necessary before re-use.

Annealing is much more complicated for some preparations of LiF and can alter the dosimetric properties of the TL material when not carried out correctly. It depends upon the form in which LiF has been manufactured. The reader is referred to the literature (Portal, 1981; Driscoll et al., 1986) and to indications provided by the manufacturers to adapt the annealing procedures to the LiF preparations used. For instance, Harshaw LiF:Mg,Ti chips or rods should be annealed from 1 to 2 hour at 400°C followed by 24 hours at 80°C after each use.

Some manufacturers recommend no annealing after re-use for the LiF detectors. In our experience correct results can be obtained provided the annealing be replaced by an adequate preheating of the irradiated material just before the readout cycle (Marinello et al., 1992).

**Practical consequences**

Li$_2$B$_4$O$_7$ either doped with Mn or Cu can be used and re-used many times without thermal treatment between the successive irradiations and readouts.

LiF, CaSO$_4$ and CaF$_2$ need specific thermal treatments after each use.
IV.3.5. Influence of dose-rate

TL dosimeters are to a large extent dose-rate independent. As shown by Tochilin (1966) and Goldstein (1972) LiF and Li₂B₄O₇:Mn are independent of the dose-rate up to 45 Gy and 10³ Gy per pulse of 0.1 μs respectively. This property implies in practice that the dose-rate variations produced by beam modifiers, SSD variations, patient thickness, etc. do not have to be taken into account with the TL method. Even the extreme high dose-rates produced in scanned electron beams of linacs do not cause any special difficulty.

Practical consequences

No correction for dose-rate is necessary in the range of clinical dose-rates applied.

IV.3.6. Influence of temperature

As the temperatures required to get the light signal out of the TL crystal is high compared to room or patient temperature, the response of TL dosimeters is independent of temperature variations in the range concerned by in vivo dosimetry. However care should be taken not to store the dosimeters close to a heat source.

Practical consequences

No correction for temperature is necessary for in vivo dosimetry.

IV.3.7. Influence of energy

IV.3.7.1. Photon energy

Except for superficial measurements, TL dosimeters should be surrounded by a suitable build-up cap corresponding to the energy and geometrical irradiation conditions considered (§ II.1). When the build-up cap is made of tissue-equivalent material, it is theoretically possible to evaluate the absorbed dose in TL dosimeters and associated build-up cap irradiated with photon beams, knowing the relative variation of mass energy absorption coefficient between the TL material considered and water as a function of photon energy. The theoretical variation is less than 4 % and 8 % in the energy range from 1 to 50 MeV for Li₂B₄O₇ and LiF, respectively. In practice, due to the influence of the surrounding material (build-up cap and patient), the size and the shape of the TL dosimeters may modify the expected results by a few per cent (Almond and Mc Cray, 1970; Chavaudra et al, 1976; Bagne, 1977). The heating conditions may also modify slightly the results (Mc Keever, 1985). So the most reliable method consists of comparing directly the TL dosimeters and associated build-up cap to be used to a calibrated ionisation chamber the energy response of
which is well known, to irradiate both the detectors in the same beam as those used for patient treatments, using the experimental conditions shown in figure III.9, and to compare their responses. Because of the slow variation of response versus energy of Li$_2$B$_4$O$_7$ and LiF in the energy range considered, the calibration factors obtained with this method can then be used for all patients treated in the same beam, or patients treated in photon beams of the same energy, irrespective of the geometrical conditions of the irradiation (field size, SSD, presence or not of compensating filters, etc).

As, for photon energies below 300 keV, TL dosimeters having to be very thin and applied without build-up cap, it is preferable to use lithium borate instead of LiF, and a fortiori other TL material (Fig.IV.7), because the variation of response with energy is less important. In this case theoretical curves of response versus energy, such as those shown in figure IV.7, can be used as a first approach every time the TL dosimeter is of small dimensions (Jayachandran, 1970). For low energies (below about 50 keV), theoretical curves or any other theoretical data, must not be used directly because the shape and dimensions of the detector can induce considerable response variations within the dosimeter volume (Greening et al, 1972; Bassi et al, 1976; Mc Kinlay, 1981). Moreover the nature of the activator (§ IV.2.1) may also yield too large differences in the response of TL materials in this energy range (Wall et al, 1982). The only solution consists of comparing directly the response of the TL dosimeters to a calibrated ionization chamber using a method similar to the one shown in figure III.9. Due to the low energy range, the effective point of measurement of the chamber, which should be adapted to these low X-ray energies, is then situated at the same level as the TL dosimeter.

![Figure IV.7: Relative response of different TL material as a function of photon energy (After Bassi et al, 1976)](image-url)
Finally it should be noted that LiF type 6 and Li$_2$B$_4$O$_7$ respond to slow neutrons via reactions with $^6$Li and $^{10}$B. As X-rays of very high energy are sometimes contaminated by neutrons, a particular attention has to be taken when in vivo measurements are performed with X-ray beams of energy greater than 12 MV. The best solution consists of using LiF enriched in $^7$Li which is not sensitive to neutrons.

**IV.3.7.2. Electron energy**

Theoretically it is possible to evaluate the absorbed dose in TL dosimeters irradiated with electron beams knowing the variation of the ratio of the mass collision stopping power of the TL material to that of tissue or water as a function of energy. This variation is less than 2% and 5% for LiF and Li$_2$B$_4$O$_7$ respectively in the energy range from 200 keV to 50 MeV (ICRU, 1984). In practice and for the same reasons as for photon beams (§ IV.3.7.1) it is preferable to compare directly the TL dosimeters to be used to a calibrated ionisation chamber the energy response of which is well known for electron beams (§IV.4). The validity of the method has been verified by different authors (Almond et al, 1967; Marinello et al, 1973; Holt et al, 1975; Bagne, 1977; Marre et al, 2000).

**Practical consequences**

The response of a TL dosimeter has to be corrected to take into account the photon or electron energy. Correction factors to be used cannot be obtained from theoretical data, except in particular conditions. They have to be determined in the same conditions as those used to treat patients and for the type of dosimeters used in practice and read out on the particular reader.

For a given TL dosimeter associated with a given reader, energy correction factors remain constant and can be introduced in the computer associated with some readout systems in order to correct the results automatically.

**IV.3.8. Directional effect**

The intrinsic response of TL dosimeters is not influenced by the direction of the beam.

**Practical consequences**

No correction for directional effect is necessary except if the dosimeter container and associated build-up cap have an asymmetric shape; even for the tangential irradiation of the breast or the thoracic wall no directional dependence of detector response is observed.
IV.4. Clinical use

Theoretically, for *in vivo* dosimetry with TL dosimeters, the same approach as for diodes could be used. Indeed, if the detectors are identified and can be kept trace of, a calibration factor could be ascribed to each TL dosimeter as it is done to a diode. It would then of course also be necessary to monitor these factors over time. However this procedure would require an extreme stability of reading, annealing and manipulation conditions, which would be very hard to achieve in practice. Moreover, keeping trace of each detector is not an easy task in case of solid detectors, whilst it is technically impossible in the case of powder.

In practice, for calibration, advantage is taken of the large number of detectors, or of the large quantity of powder available. In this respect it is perfectly possible to save some detectors (or an amount of the powder) for the purpose of calibration and to convert the reading of the patient detectors to dose by comparison of their signal with the calibrated detectors. Instead of having to monitor calibration factors over time, at each readout session patient detectors are just analysed together with some calibration detectors. These methods, applied in TLD, are presently well documented and, thanks to the availability of modern automated readers, the large scale applicability of this dosimetry method, especially for *in vivo* measurements, has increased considerably.

**IV.4.1. Calibration of TL dosimeters**

Basically the methods for calibration of TL dosimeters are very similar to those for diodes (§ III.4.1). The same phantom set-up (Fig.III.9) can be used and again the ionisation chamber will be used as reference instrument. Entrance as well as exit dose calibration factors may be determined. However, the only factor of importance for TL dosimeters being the difference in relative distances from ionisation chamber and detector to the source, the ratio F (§ III.4.1.2) between the 2 factors will be much closer to unity than for some diodes.

It is not always necessary to irradiate new calibration detectors for each read-out session. Calibration TLD, having been irradiated together, may be attributed to different sessions as long as the time gap between their irradiation and that of the patient detectors is short enough in order to limit fading phenomena to less than 1% (Table IV.3).

**IV.4.2. Use of TL dosimeters in clinical conditions**

Every factor which influences TL response (§IV.3) may be the reason why, when the clinical conditions are different from the reference conditions, correction factors may be necessary. In fact, provided the calibration of the detectors be performed with a beam of the same radiation quality as the one used to treat patients, no correction factor has to be used except a correction factor, \( C_D \), for non linearity of dose response with some TL materials (Table III.2). This makes TL detectors particularly attractive for *in vivo* dosimetry, including in vivo dosimetry in IMRT (Burman et al, 1997; Van Esch et al, 2002; Engström et al, 2005).
However, as for diode dosimeters and for the same reason, this type of application falls outside the scope of present work.
V. DIODE OR RADIOTHERMOLUMINESCENT DOSIMETERS: WHICH METHOD?

The choice for a radiotherapy department between diodes and TLD may depend on many factors like intrinsic characteristics of the detector type, environment in personnel (e.g. size of the physics group), financial considerations and, of course, personal preference. Often the decision is not “black or white”, and only in a few cases major arguments in favour of one or the other method exist. Furthermore, as seen (chapters III and IV), diodes and TLD, when correctly chosen and handled, will both lead to a very good accuracy.

With respect to diode dosimetry, it is obviously the only one offering the possibility for immediate information, with the patient still on the treatment couch. When a deviation is observed a visual check of the field set-up may then be performed without delay. But full exploitation of this attractive feature might be less straightforward than expected, and in any case, requires a well adapted flow chart of necessary preparatory actions and of different evaluations and corrective actions, which can take time.

Diodes are also very convenient when, in the case of parallel-opposed beams, the entrance and exit doses have to be determined for each beam separately. Indeed, provided the entrance and exit dose calibration factors of both diodes have been measured, a reading of their signal after each irradiation is the only requirement, and no change of detector is necessary in between.

Thanks to the modern automated readers TLD results may now be obtained within some 15 minutes after irradiation. This method is indicated in different situations in external radiotherapy (its obvious advantages for low dose-rate intracavitary measurements falls outside the scope of this work). Even with the multichannel electrometers presently available for diode dosimetry, the number of sites to be explored may sometimes become too large (e.g. for dose evaluation at different places in large fields as applied in TBI or total skin irradiation). Thanks to the very high number of detectors available and to the absence of cables TLD offers attractive possibilities.

Due to the limited size of the detectors and therefore their excellent spatial resolution, another typical application of TLD is the exploration of zones of possibly high dose gradient. It is then possible to study e.g. field junction zones by juxtaposing on the patient’s skin a number of TLD chips, or a number of small containers to be filled with TLD powder. Dose distributions at the entrance or exit surfaces of the 2 adjacent fields may then be derived with a high spatial resolution.

Thanks to their tissue-equivalence TLD are also of interest for measurements outside the field, because, despite the spectral variations with in-field positions, the calibration factor derived at the beam axis may still be used with a reasonable accuracy. They can also be used with low energy X-rays which is not the case with diodes because of their high atomic number (§ III.3.7.1).

Apart from these few examples, where some preference exists for one or the other method, the choice is much less straightforward in most of the other applications. A
comparison of the intrinsic characteristics and of the organisational aspects of both methods could provide some useful information.

V.1. Comparison of intrinsic characteristics of TLD and diodes.

The main characteristics, as presented in § III.3. and § IV.3., are summarized in Table V.1. For comparison the characteristics of the ionization chamber have also been included. In summary, diodes score better than TL dosimeters with respect to immediate response, share the advantage of absence of high voltage, but are inferior with respect to presence of a cable, response variation as a function of accumulated dose, dose-rate, temperature, energy (provided the TL material is well adapted (Table IV.1)) and also with respect to directional effects (Lagrange et al, 1988). However, for exit dose measurements it has been shown (Aukett, 1991) that, for 6-MV X rays, the variations in spectrum with depth are not significant either with diodes or TL dosimeters. Furthermore, the importance of each factor has obviously to be evaluated in the context of the users’ needs.

Table V.1: Principal advantages and disadvantages of diodes and RTL detectors for applications in in vivo dosimetry. The characteristics of the ionisation chamber have been shown for comparison.

<table>
<thead>
<tr>
<th></th>
<th>Cable delay</th>
<th>High voltage</th>
<th>Delay in results</th>
<th>Dose</th>
<th>Dose accum.</th>
<th>Dose-rate</th>
<th>Temperature</th>
<th>Energy</th>
<th>Direction</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ionisation chamber</td>
<td>XX</td>
<td>XX</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>X</td>
<td>X</td>
<td>X</td>
<td>0</td>
</tr>
<tr>
<td>Diode</td>
<td>X</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>XX</td>
<td>XX</td>
<td>X</td>
<td>XX</td>
<td>X</td>
</tr>
<tr>
<td>Thermoluminescent</td>
<td>0</td>
<td>0</td>
<td>X</td>
<td>0 or</td>
<td>X</td>
<td>0</td>
<td>0</td>
<td>X</td>
<td>0</td>
</tr>
</tbody>
</table>

0 no concern       * depending on the TL material and associated reader
X mild concern
XX serious concern

V.2. Organisational aspects of in vivo dosimetry

V.2.1. Diode dosimeters.

To take full advantage of the possibility of diode dosimeter for immediate information, it is mandatory to know the expected detector signal at the very moment of the in vivo dosimetry. The preliminary calculations of the expected signal have to be done by the physics team before the irradiation.

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The work around the patient, at the moment of measurement, is relatively extensive and complicated. The critical point is the comparison of the measured detector signal with the expected one: if the difference exceeds a predetermined action level (e.g. +/-5%), the reason must be investigated without any delay and very often this will imply visual check of the set up parameters (SSD, collimator opening, presence of beam accessories, etc) near the patient, who is still in the treatment position. A major practical issue is then obviously related to the workload and responsibilities at this crucial step of the procedure. There is little doubt that radiotherapy technicians can be trained to position correctly the detectors on the patient’s skin, to operate the electrometer and to read its signal. With respect to this last step, it is recommended to have an automated independent recording system, in order to allow a posteriori check of the results.

It should be no problem for the technicians to judge whether the deviation between measured and expected signal exceeds the action level agreed upon in the department. In the case of observed deviations they can also perform, with the patient still on the treatment couch, a first (quick) check of the irradiation parameters and inform the staff about their findings. If the deviation remains unexplained, a new measurement is to be performed at the next session, however this time in the presence of a physicist or a radiotherapist.

V.2.2. Radiothermoluminescent dosimeters

The main practical difference between TLD and diodes is that the work intensive periods occur at different stages of the radiotherapeutic process. While for diodes the major part of the effort is to be delivered at the moment of treatment itself, the evaluation of TL detectors is performed at a later stage. As such it is important to realise that the TL method will modify the overall clinical procedure much less than diodes (if, at least, these are used to collect on-the-spot information). With TLD the main workload of course occurs at readout of the samples. However, thanks to the increasing availability of automated TLD readers the amount of working time required for readout has decreased substantially. However any investigations to find out the reason for a deviation can be more difficult than with diodes because of the delay, resulting in an additional workload.

Because the number of TL detectors to be evaluated is usually relatively limited in a radiotherapy department (except when special techniques such as total body or total skin irradiation are practiced), the price for such an automated reader may seem at first sight prohibitive. A pragmatic solution to the financial issue could be the sharing of the equipment with the radiation protection department. Finally the reusability of the TL dosimeters is also an important aspect of the financial issue. When taking into account the total annual costs, the TLD system has been estimated to be cheaper than diodes only for very small and for very large departments, and this for a semi-automatic and automatic reader (Kesteloot et al, 1993).
VI. ALTERNATIVE METHODS AND FUTURE DEVELOPMENTS

As shown in chapters III to V, diodes and thermoluminescent dosimetry are the two well-established methods used for in vivo dosimetry. There are however a number of alternative methods which could present their own typical advantages in the near future. Amongst these methods Metal Oxide Semiconductor Field Effect Transistors or MOSFET dosimeters deserve special attention because they are used in a growing number of radiotherapy centres. Although two other methods, plastic scintillators and diamond detectors, present promising characteristics for in vivo dosimetry, there is still a striking lack of breakthrough for this type of application. Another method, making use of radiochromic films, has in recent years been used on patients undergoing high-dose interventional radiological and brachytherapy procedures and presents also interesting possibilities for use in external radiotherapy. Optically Stimulated Luminescence (OSL), with already a sound base for a range of dosimetric applications, is presently the subject of research for extension of its field of application to in vivo dosimetry. Finally, in Electron Paramagnetic Resonance (EPR), the classical molecule alanine is being replaced by other alternatives more suited for in vivo dosimetry in external radiotherapy.

VI.1. MOSFET dosimeters

Like diode dosimeters MOSFET dosimeters belong to the category of the semiconductor detectors. The MOSFET (Fig. VI.1) - a p-type in the case shown on the figure – is built on a silicon substrate of the opposite sign (in this case thus negatively doped). Two terminals of the MOSFET called the source and the drain are situated on top of a p-doped silicon region. The third terminal shown is the gate. Underneath the gate is an insulating silicon dioxide layer and underneath this layer is the silicon substrate. The region of the substrate immediately below the oxide layer is known as the channel region.

![Schematic representation of a p-channel MOSFET showing the oxide, the substrate, the source, the drain and the gate](image)

Figure VI.1: Schematic representation of a p-channel MOSFET showing the oxide, the substrate, the source, the drain and the gate (After Soubra et al, 1994).

1 The authors are indebted to J. Cygler, J. Gray and E.S. Bergstrand for providing substantial information about MOSFETs, OSL and EPR, respectively
When a bias voltage is applied between source and drain, a sufficiently negative voltage – the “threshold voltage” $V_{TH}$ – must be applied at the gate in order to allow a predetermined current $I_{ds}$ to flow. The underlying mechanism is that the negative gate voltage causes holes from both the bulk of the silicon to move close to the oxide-silicon surface. Once a sufficient number of holes accumulated there, a conduction channel is formed, allowing the current $I_{ds}$ to flow between source and drain. During the irradiation of the MOSFET a number of the holes produced are captured into traps close to the Si-SiO$_2$ interface. During the irradiation the gate is kept at a positive bias in order to enhance this phenomenon and thus to increase the sensitivity. The trapped holes cause a negative voltage shift $\Delta V_{TH}$ (Fig. VI.2) in the threshold voltage, which is proportional to the number of trapped holes. This voltage shift can then be used for dosimetry. Before and after the irradiation the threshold voltage must be measured.

Several authors (Soubra et al, 1994; Gladstone et al, 1994; Butson et al, 1996; Ramani et al, 1997; Scalchi and Francescon, 1998; Chuang et al, 2002-a; Ramaseshan et al, 2004; Halvorsen 2005) have determined their characteristics as in vivo dosimeters. Some results obtained on patients or on phantoms have also been published (Cygler et al, 1995; Soubra et al, 1996; Quach et al, 2000; Chuang et al, 2002-a; Bloemen-Van Gurp, 2003; Jornet et al, 2004; Marcié et al, 2005; Scalchi et al, 2005).

Their advantages are:

- “immediate” response; however not really “on-line”, because after termination of the irradiation the gate voltage $V_{TH}$ must be determined
- No need for cables to connect the detectors on the patient to an electrometer; however the gate must then be prebiassed before the irradiation
- small size (of the order of 1 mm$^3$)
- good reproducibility: 2-3% (1 $\sigma$)
• non-destructive read-out: possibility of study of dose accumulation over the different treatment sessions, however with a fading correction
• response independent on dose-rate
• negligible angular dependence (+/- 2\% for 360\°)

Their disadvantages are:
• response dependent on temperature except for dual base dosimeters; their principle consists of associating 2 detectors with a different grid voltage and of displaying the difference in their signals in order to produce temperature compensated dosimeters.
• response decrease as a function of accumulated dose, resulting in a limited lifetime of the detector (of the order of 10^2 Gy in the usual conditions)
• Response dependent on energy: as for diodes, the basic material of a MOSFET is silicon so that a similar energy dependence is observed
• slight loss of charge after irradiation: readings to be taken always at the same time delay after termination of the irradiation

VI.2. Plastic scintillator dosimeters

Potential uses of solid organic scintillators were first mentioned by O’Foghludha in 1969 (O’Foghludha, 1969) but interest in their applications in medical radiation dosimetry has only been shown in the 90’s (Beddar et al, 1992-a and b; Perera et al, 1992; Frelin et al, 2005). The scintillator probe consists of a base material (usually polystyrene, polyvinyl-toluene or acrylic-naphtalene) and one or more organic dyes embedded in it. When photons or charged particles pass through it, a few per cent of the energy loss is ultimately converted to scintillation photons. The light intensity is then either amplified and collected by photomultiplier tubes (Beddar et al, 1992-a), or focussed on the photocathode of an image intensifier by a light guide and captured by a charge coupled digital camera (Perera et al, 1992). In both the cases the plastic detector yields a count rate which can be correlated to the absorbed dose received by it.

Water-equivalence for plastic scintillators can be achieved by a good choice of the compound materials and become superior to the majority of other detector types (Clift et al, 2000). As for their other main properties their excellent reproducibility and stability, their negligible dependence on temperature, their linear response versus dose from some cGy to some Gy, their independence of dose-rate and their high spatial resolution must be mentioned. Due to these advantages, plastic scintillator detectors would require only a very limited number of correction factors when used for in vivo dosimetry. Unfortunately the dosimetric systems currently in existence, developed to establish dose distribution around radioactive sources or within photon or electron beams, are not yet optimized for in vivo measurements (De Boer et al, 1993), although the development of multi-channel PMTs may offer improved possibilities in this respect (Flühs et al, 1996; Létourneau et al, 1999).
VI.3. Diamond dosimeters

The potential use of natural diamonds as conduction-type radiation counters were mentioned by Cotty as early as 1956 (Burgemeister et al., 1981). Moreover, with a low voltage applied to the diamond, it was found to operate as a radiosensitive resistor. More recently diamonds have been used as solid-state ionization chambers. The radiation produces electron-hole charges in the crystal, to which an external bias of typically 100 V is applied to separate the charge carriers (Schüle and Hodapp, 1992). The resulting current is measured with an electrometer. After the external bias is switched on, the probe must be irradiated to about 5 Gy in order to prefill the crystal lattice electron traps. The detector response then remains stable until the bias is switched off.

The superior intrinsic characteristics of diamond detectors for dosimetry have been illustrated (Vatnitsky et al., 1993-a and b) by showing that, for determination of absorbed dose to water in a phantom, the possibilities offered by this type of detector are comparable to those of ionization chambers. With respect to in vivo dosimetry, diamond probes seem to share most of the attractive characteristics of plastic scintillators (§ VI.2) i.e. a good water-equivalence thanks to an atomic number of 6, a very low dependence on accumulated dose as well as on temperature, and a very impressive spatial resolution. On the other hand, they present the inconvenience of a dose-rate dependence more pronounced than for scintillators or diodes (§ III.3.5). Even with cobalt 60, a dose-rate dependence has been observed at dose-rate > 5 Gy/min for a bias of 100 V. Obviously this dose-rate effect calls for caution when using diamond detectors with linacs. Moreover, it has been shown that they present a directional effect when used with electron beams (Heydarian et al., 1993) As for scintillators, the use of diamonds for in vivo dosimetry is presently not yet established and needs further efforts.

VI.4. Radiochromic film dosimeters

First proposed by Mc Laughlin and Chalkley (1965), the use of radiochromic films for in vivo dosimetry is relatively recent. They have been used for in vivo dosimetry performed on patients treated with kilovoltage (Haque et al., 2003) or high energy X-rays (Cheung et al., 2002-b) and with intraoperative electron beams (Ciocca et al., 2003). For measurement of the entrance dose in high-energy photon beams, films have to be covered with some build-up sheet (Quach et al., 2000), except when surface or skin doses have to be determined (Butson et al., 1998-b and 1999). The major advantage of this type of detector is the possibility to obtain complete dose distributions with high spatial resolution (Dempsey et al., 2000). This is particularly important with respect to the dose at field and block edges, at tissue interfaces, in small fields (Yamauchi et al., 2004) and IMRT fields.

The present possibilities offered by radiochromic films are essentially due to the improvement of the homogeneity and sensitivity of commercial emulsions (Meigooni et al,
They consist of one or several sensitive layers made of microcrystals of radiation-sensitive monomers uniformly dispersed in a gelatin binder. Films are colorless and transparent before being irradiated. When they are exposed to ionizing radiation, polymerisation of the microcrystals occurs and the polymers alters the crystal colour to various shades of colour depending on the type of film (e.g. blue for Gafchromic films type MD 55, HS and EBT, and orange for type XR-T or RTQA). The mechanism for radiation induced coloration has been demonstrated to be essentially due to a first order solid state polymerisation of the ordered substituted diacetylene monomers, which produce a main carbon chain planar conjugation (Mc Laughlin et al 1996).

The colour change depends on the absorbed dose and can be measured with any densitometers, flatbed color scanners or spectrometers but the measurement is optimised through the use of instruments specially designed to measure the principal absorption peak in the spectrum of the used Gafchromic film (AAPM, 1998; Gluckmann and Reinstein, 2002; Chui-Tsao, 2004; Devic et al, 2004; Fusi et al, 2004). A small variation of wavelength can have an important impact on the results (Sullivan et al, 2000). For instance reading out the blue coloration of Gafchromic films MD 55-2 or HS with specially designed densitometers increase the sensitivity by a factor 2 or 3 compared to black and white densitometers or He-Ne laser scanners. Results obtained using a black and white densitometer can be significantly increased by using a red or yellow filter while making densitometer measurements. When films are read out with a medical film scanner using a white light source (for instance VIDAR scanners), the response can be improved by placing the film behind a red or yellow filter sheet but this leads to a small reduction in the light intensity, and so the resulting measurements have a lowest signal-to-noise ratio. Modern document scanners associated to commercial software can also give acceptable results at less expensive cost (Stevens et al, 1996; Aydarous et al, 2001; Alva et al, 2002; Yamauchi et al, 2004).

During the past several years the production process for these films was modified to enhance their sensitivity, homogeneity and reproducibility. These modifications included changing the chemical composition of the emulsion, changing the number and the thickness of the sensitive layer, and modifying the technique of distributing the sensitive material on the film base. As conditions of readout and properties of radiochromic films are highly dependent on the previous parameters, it is highly recommended to look at both the type of film and the batch number, and to check its dosimetric properties on its own equipment, before using it.

The advantages of radiochromic films are summarized below. They suppose that the film digitizer is adapted to the type of films used and that its performances are regularly verified (Meeder et al, 1995; AAPM, 1998):

Details on composition and dosimetric properties of Gafchromic films can be found on: www.ispcorp.com/products/dosimetry/index.html
– direct imaging and self-developing (no physical, chemical or thermal processing), so variations introduced by the processing step are eliminated.
– easily handled under incandescent light (Butson et al, 1998-a), but care must be taken when used under long wavelength UV light which may induce polymerisation without ionizing radiation
– very thin and easily cut to the desired shape and dimensions (films of dimensions up to 20 x 25 cm² are now available)
– water resistant (Butson et al. 2001) good reproducibility from 2 to 5% (1 σ) depending on the type of film, readout system and range of doses, provided a constant delay between irradiation and readouts is maintained, or a delay varying from 2 up to 24 hours or more, depending on the type of film and associated readout system and dose, be respected between irradiation and film readout (Klassen et al, 1997)
– non-destructive readout which offers the possibility of dose accumulation over the different treatment sessions provided correction factors be introduced to take into account the variation of optical density with time
– high spatial resolution
– approximately tissue-equivalent over the range of photon and electron energies used in radiotherapy (Mc Laughlin et al, 1991) except at very low photon energy.
– response independent on dose-rate from 0.08 to 80 Gy/min (Mc Laughlin et al, 1996). However, for exposures over long time period, or for exposures that cause only small changes in net density, Ali et al. (2003) have reported some dependence of the film response on dose and dose fractionation.

The disadvantages of radiochromic films are:
– reduced sensitivity compared to silver based films: for instance the response range for Gafchromic HS is 0.5-40 Gy but is improved for Gafchromic EBT recently introduced on market which can be used in the range 0.2 Gy - 40 Gy
– dose response curve slightly curved with reduced sensitivity at higher doses which implies corrections. However, it should be noted that such corrections are easy to be taken into account in modern readout systems.
– several hours are necessary for the colour change to be sufficiently stabilized for evaluation (Klassen et al, 1997, Ali et al, 2003 ). In practice it is recommended to read out the films at constant delay after irradiation or to wait for a sufficient delay before readouts to get reproducible measurements. The variation of optical density with time is also temperature dependent so that care must be taken for film storage. For instance Reinstein and Gluckman (1999) have shown a variation of 4% between 20 and 35°C for Gafchromic MD 55-2 films. The measured optical density in Gafchromic MD 55-2 have been shown to be dependent on the polarization of the light used for evaluation by Klassen et al (1997) . They recommend that orientation and alignment of the film be identical for calibration and measurements. New types of films seem to be less sensitive to that problem. Recently Fusi et al. (2004) have investigated the possible causes of inaccuracy and shown that they were
highly dependent on the type of film and commercial readout system used because the scattering coefficient of Gafchromic films is not negligible.

VI.5 Optically stimulated luminescence dosimeters

Optically stimulated luminescence (OSL) dosimetry is very similar to TLD because the same basic mechanism is involved, however with stimulation by light instead of heat. The future of OSL dosimetry as an alternative to the well established TLD method is at present a matter of debate (McKeever, 2002; McKeever and Moscovitch, 2003). Advantages of OSL dosimeters which are advocated are an increased sensitivity and the possibility of doing repeated read-outs. Moreover not only the laser stimulated emission after the irradiation, proportional to the absorbed dose, but also the prompt radioluminescent (RL) emission observed during the irradiation, proportional to the dose-rate, can be used for dosimetric purposes.

Several materials (some exhibiting also TLD properties) can be used for OSL dosimetry purposes. In the past silica glasses have been used for personal and medical applications (Polf et al, 2002). The industrial (Landauer Inc, Chicago, USA) development of the carbon-doped aluminium oxide (Al$_2$O$_3$:C) radiation sensitive saphire has led to the development as a luminescent dosimetry material with applications in environmental, personal and space dosimetry. Recently there has been a growing interest for the potentiality of OSL for in vivo dosimetry. The characteristics of the Al$_2$O$_3$:C dosimeter have been explored for this type of application. The response has been found linear as a function of dose between 0.05 and 100 Gy (Polf et al, 2002) and between 0 and 3 Gy (Andersen et al, 2002). It has also been found to be independent within 1% of dose-rate between 0.85 and 5.1 Gy/min (Aznar et al, 2004) and between 0.8 and 3.1 Gy/min (Andersen et al, 2002). As pointed out by Aznar et al (2004), OSL response may vary with the radiation quality used, due to its relatively high Z (10.2). However only a limited (within 1%) energy dependence has been observed between 6 MV X-rays, 18 MV X-rays (Aznar et al, 2004; Andersen et al, 2002) and 20 MeV electrons (Andersen et al, 2002). Although further research on their basic characteristics is needed, OSL dosimeters have already been used on patients (Meeks et al, 2002; Aznar et al, 2004).

VI.6. EPR dosimeters

EPR, or Electron Paramagnetic Resonance, is a magnetic phenomenon with a lot of applications in physics, chemistry and other disciplines. Also in medical and health physics dosimetry it has a number of interesting possibilities.

The interaction of ionizing radiation with matter results in the formation of free radicals, that is, molecules having one or more unpaired electrons. With EPR spectroscopy, the induced transitions in the electron spin states may be used to obtain the relative amounts
of free radicals in a sample (Box and Freund, 1959). The most common EPR dosimeter material is the amino acid L-α alanine, and the associated EPR signal is proportional to the dose over a wide dose range (1-5000 Gy). The effective atomic number of alanine is close to that of water, and its response (readout per dose to water) is nearly independent of the photon and electron beam qualities commonly used in radiotherapy (Olsen et al, 1990). The readout procedure is non-destructive to the EPR signal. For alanine, the EPR signal is quite stable; reported signal fading rates are about 4% per year when the dosimeter is stored under “normal” laboratory conditions (Sleptchonok et al, 2000). The dependence of the EPR signal on influence factors such as air humidity and irradiation temperature is low (Nagy et al, 2000). Any dose-rate dependence is not detected for dose-rates up to $10^5$ Gy/s and $10^8$ Gy/s for continuous and pulsed radiations, respectively (McLaughlin et al, 1989; Regulla and Deffner, 1982).

EPR/alanine dosimetry is currently used for mailed dose calibration services at several dosimetry laboratories worldwide. Dedicated benchtop spectrometers with automated readout are available. Compared to TLD, the sensitivity is less for doses below about 1 Gy. Improved precision and accuracy in the subgray range is possible, but requires either rather large dosimeters or elaborate readout regimes (Hayes et al, 2000). Hence, the current system does not immediately apply to measure doses in vivo in fractionated radiotherapy (in the order of 1 Gy) with accuracies better than about 4-5%, using a realistic amount of readout time (Ciesielski et al, 2003; Nagy et al, 2002).

An ideal EPR dosimeter material for the radiotherapy dose range should be more radiosensitive than alanine, while possessing similar favourable characteristics as this material. There are ongoing investigations of more sensitive alternative materials. One example is formates and dithionates that are 6-7 times more sensitive than alanine (Lund et al., 2004; Vestad et al, 2003), while having many of the desired characteristics for good in vivo dosimeters. It may be expected that EPR dosimetry will become a useful alternative for in vivo dosimetry, provided that the sensitivity to dose is increased by the use of suitable dosimeter materials.


