Investigating diode detectors for in vivo dosimetry in intensity modulated radiation therapy

Nils Kadesjö

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Thesis for Master of Science in Medical Radiation Physics
Supervisor: Jörgen Olofsson
Assistant Supervisor: Göran Rikner
Examiner: Heikki Tölli

Umeå University
Department of Radiation Sciences
Radiation Physics
SE-901 87 UMEÅ
SWEDEN
Abstract

In radiotherapy accurate delivery of the prescribed dose is vital. One way of verifying the delivered dose is through in vivo dosimetry, i.e. measuring the patient dose during treatment. According to Swedish regulations, in vivo measurements is required for each patient undergoing external beam radiotherapy. Silicon diode detectors have been used for in vivo dosimetry in conventional radiotherapy for many years. For complicated techniques like intensity modulated radiation therapy (IMRT), no routine in vivo measurements are regularly performed due to difficulties in measuring the dose accurately. This work investigates the feasibility of performing routine in vivo measurements for static multi leaf collimator-IMRT with the aid of the quality assurance software EqualDose.

Two diode detectors from IBA Dosimetry have been studied: the hemispherical EDP-10 and the cylindrical RDP. 6MV photons were used for all tests. The effects of varying field sizes, source-to-surface distance, temperature, angle of incidence and off-axis locations have been studied. Both phantom and in vivo measurements were performed for IMRT in the head and neck region.

For EDP-10 the IMRT measurements showed a mean deviation from calculated value (with standard deviation in parenthesis) of 2.0(1.3)% using a flat phantom, 1.4(2.0)% with cylindrical phantom and 0.4(3.0)% for actual in vivo measurements. The corresponding deviations for RDP were 1.9(1.8)%, 2.0(2.1)% and 3.7(2.2)%, respectively.

From this work it can be concluded that routine in vivo dosimetry for IMRT, using a 5% tolerance level, is feasible.
Abstract


IMRT-mätningarna med EDP-10 gav följande medelavvikelse från förväntat värde (med standardavvikelsen innom parentes): 2.0(1.3)% med platt fantom, 1.4(2.0)% med cylindriskt fantom och 0.4(3.0)% för verkliga in vivo-mätningar. Motsvarande mätningar med RDP gav följande avvikelser: 1.9(1.8)%, 2.0(2.1)% samt 3.7(2.2)%.

Av denna undersökning kan slutsatsen dras att rutinmässiga in vivo-mätningar med en toleransnivå på 5% är genomförbara, också vid IMRT-behandlingar.
Contents

1 Introduction

1.1 External beam radiotherapy ................................................. 2
  1.1.1 IMRT ................................................................. 3

1.2 In vivo dosimetry .............................................................. 3
  1.2.1 Diode dosimetry ..................................................... 4

1.3 EqualDose ................................................................. 4

1.4 Aim of the report ............................................................. 6

2 Materials and methods .......................................................... 7

2.1 Equipment ............................................................................ 7
  2.1.1 Detectors ................................................................. 7
  2.1.2 Accelerator ............................................................... 7

2.2 Basic detector characteristics .................................................. 7
  2.2.1 Field size dependence .................................................. 7
  2.2.2 SSD dependence ......................................................... 8
  2.2.3 Angular dependence ...................................................... 8
  2.2.4 Temperature dependence ................................................. 8
  2.2.5 Off-axis dependence ...................................................... 8

2.3 IMRT measurements ............................................................. 9
  2.3.1 IMRT using a flat phantom .............................................. 9
  2.3.2 IMRT using a cylindrical phantom .................................... 9
  2.3.3 IMRT verification in vivo ............................................... 9

3 Results .................................................................................... 10

3.1 Basic detector characteristics ................................................... 10
  3.1.1 Field size dependence .................................................. 10
  3.1.2 SSD dependence ......................................................... 11
  3.1.3 Angular dependence ...................................................... 12
  3.1.4 Temperature dependence ................................................. 13
  3.1.5 Off-axis dependence ...................................................... 14

3.2 IMRT measurements ............................................................. 15
  3.2.1 IMRT using a flat phantom .............................................. 15
  3.2.2 IMRT using a cylindrical phantom .................................... 15
  3.2.3 IMRT verification in vivo ............................................... 16
Glossary

3D-CRT Three dimensional Conformal Radiotherapy
CT Computed Tomography
IMRT Intensity Modulated Radiotherapy
MLC Multi Leaf Collimator
SSD Source-to-Surface Distance
SVWT Sensitivity Variation With Temperature
TPR Tissue to Phantom Ratio
Chapter 1

Introduction

Treatment of cancer through radiotherapy dates back to the start of the 20th century. While the methods for radiotherapy have improved greatly since those early days the basic principle remains the same. Through the energy deposited by ionizing radiation (the energy deposited to a point per mass is known as the absorbed dose (with the unit J/kg or gray)) the DNA of the cancer cells are damaged, preventing the cells from proliferating. The DNA of healthy cells are also damaged by the radiation but as they have a greater ability to repair themselves, this generally results in less damage to healthy tissue than tumour tissue at the same absorbed dose. The objective of radiotherapy is to produce a sufficiently high dose to tumour tissue, while at the same time reducing the dose to healthy tissue. Relatively small deviations in patient dose may have serious consequences. A too low dose can reduce the biological effect to tumour tissue and a too high dose to healthy tissue can result in unnecessary complications. This makes quality assurance an important aspect during radiotherapy.

1.1 External beam radiotherapy

In external beam radiotherapy the type of radiation most commonly used consists of high energy photons. Historically the radiation was usually obtained from radioactive sources but modern systems generally use linear accelerators. In the accelerator electrons are accelerated to the desired energy, typically between 4 and 20 MeV, and then collided with a high atomic number x-ray target, producing a photon spectrum.

Historically the planning of radiation treatments have been performed with the aid of two-dimensional x-ray images. The treatments where delivered using collimator blocks to produce rectangular fields, covering the desired treatment volume. In order to better conform the field to the target volume, thus reducing the dose to healthy tissue, individually made shaping blocks were often used.

Advances in medical technology, particularly the introduction of computed tomography (CT) and multi leaf collimators (MLCs) combined with advancement in computer hardware and software, have allowed for improvements in planning and delivery of radiotherapy in the shape of three-dimensional conformal radiotherapy (3D-CRT). Through the use of three-dimensional anatomical data from CT images three-dimensional planning systems allow for better target definition and dose calculation, reducing the dose to surrounding normal tissue. Each field is shaped by the MLC (see figure 1.1) allowing good conformity to the target volume without the time consuming use of individual shaping blocks. The three-dimensional dose distribution is calculated after all beam parameters have been defined, a method called forward planning. Treatments using the above approach with forward planning and uniform fluence within each beam are often called conventional 3D-CRT or simply conventional radiotherapy in order to distinguish from intensity modulated radiotherapy (IMRT) treatments.

Figure 1.1: The leaves of an MLC conforming the radiation field to the treatment volume during 3D-CRT.
1.1.1 IMRT

IMRT differs from conventional 3D-CRT in two ways: non-uniform radiation beam intensities and dose planning using computer based optimization techniques.[2] The laterally non-uniform intensities are typically used to reduce the dose in particularly sensitive structures,[3] but also to compensate for irregularities in patient geometry like in the head and neck region. The optimized dose distributions are made possible by the synergistic effects of multiple beams and the computer based optimization techniques used in IMRT planning. There are two main methods for delivering the non-uniform beam, segmental multileaf collimator (SMLC)-IMRT and dynamic multileaf collimator (DMLC)-IMRT.

In SMLC-IMRT, usually called step-and-shot or similar names, the radiation is only turned on while the MLC leaves are stationary. Each beam consists of several MLC segments which together form the desired intensity distribution.

In DMLC-IMRT the MLC leaves are moving during the radiation delivery. Sliding window is a common technique where the opening formed by each leaf pair slides across the field, forming the intensity distribution.

Treatment planning for IMRT differs from conventional 3D-CRT in that it utilizes inverse planning. In inverse planning the clinical objectives, e.g. the prescribed dose to the treatment volume and the maximum dose to organs at risk, are specified and an optimization algorithm automatically determines the beam parameters.

1.2 In vivo dosimetry

In vivo is Latin for within the living and denotes the use of a whole, living organism. In clinical dosimetry this means the measurement of dose received by the patient during treatment, as opposed to dose measurements before or after the treatment, using a phantom to represent the patient. In vivo dosimetry is the most direct way to assess the dose delivered to the patient and is therefore used as an additional safeguard to verify that the treatment was delivered correctly. This complements pre-treatment quality assurance (QA) methods where dose calculation and data transfer are independently checked.[4, 5] It has been recommended that deviations of more than 5% from the expected dose should be investigated.[6] In vivo measurements are generally performed once for each patient, followed by additional measurements only if needed. Another possible approach is to perform measurements at regular intervals during the treatment series (e.g. once a week). In vivo measurements have rarely been used at each fraction of the treatment since this would result in increased treatment times and workload. Additionally, the detectors will to some extent absorb the radiation beam resulting in a dose perturbation directly below the detector. It has been suggested that due to dose perturbation, it is unsuitable to perform in vivo measurements in more than 20% of the fractions when using diode detectors.[7] Routine performance of in vivo measurements at an early stage of treatment can detect systematic errors which might otherwise accumulate along the entire course of the treatment, resulting in considerable over- or underexposure.[8] However, unsystematic treatment errors may also occur and remain undetected since no in vivo measurement was performed during that particular fraction. But if the error is a one time event, it will generally have a very small impact on the total treatment dose.

Most in vivo measurements employ either of the two well established methods, diodes and thermoluminescent dosimetry (TLD).[9, 5] In both methods the detector is normally placed on the patient’s skin and the dose to a point inside the patient is deduced from the detector reading. While diodes are popular due to their ease of reading, high sensitivity and sturdiness, the readout is not directly related to the dose.[4] In order to obtain an accurate estimate of the dose the use of several correction factors is needed. A complication encountered in photon treatments is electron contamination, i.e. the presence of unwanted electrons in the beam.[10] These electrons are produced by photon interactions and is always present to some extent. The dose contribution from electron contamination varies with field sizes and distance, leading to greater uncertainties at small depths. Higher photon energies are associated with more electron contamination which penetrates deeper into the patient then for lower energies.

For IMRT in vivo measurements become more complicated since the fluence varies within each field. This may lead to large uncertainties due to high dose gradients at the field center. Additional complications appear for some fields where there is a low dose area at the central axis, making off-axis placement of the diode necessary. These low dose areas are common in head and neck cancer when the isocenter is placed inside the medulla. According to Swedish regulations it is mandatory to perform in vivo measurements the first time a new field is delivered to the patient.[11] For conventional treatments diode detectors are usually used but for IMRT no in vivo measurements have previously been performed, in spite of the regulations. At the University hospital of Umeå in vivo dosimetry is currently being implemented for prostate and head and neck treatments.
1.2.1 Diode dosimetry

The construction of silicon diode detectors for radiation detection have been been discussed in the literature.[12, 13, 14, 15, 16, 17, 18, 19, 20] They consist of a diode chip, usually located inside a build-up cap, connected to an output cable. The build-up cap is used to emulate the measurement at a certain depth and to reduce the effect of electron contamination. A part of the diode chip makes up the effective ionization volume. The energy imparted inside the ionization volume will result in an electric current. This electric signal is measured by an electrometer and can be related to absorbed dose. There are, however, several factors that affect the diodes dose response, namely the influence of: dose rate, accumulated dose, temperature, energy, incident beam direction.[9] Due to these basic properties of diode detectors several parameters must be taken into account when performing in vivo dosimetry.

The source-to-surface distance (SSD) is the distance between the x-ray target and the patient surface. Since the x-ray target is much smaller than the SSD the target can be seen as a point source. This means that the intensity will decrease according to the inverse-square law.

\[ \frac{I_1}{I_2} = \left( \frac{r_1}{r_2} \right)^{-2} \]  

where \( r \) is the distance to the source and \( I \) is the intensity.

Diodes show a SSD dependence slightly different from the inverse-square law. This is mainly due to dose rate (silicon diodes have a superlinear response to dose rate) and the electron contamination.[21] In order to minimize the effect from the SSD dependence diodes should be calibrated at an SSD corresponding to a typical treatment distance. With low SSD dependence and suitable calibration distance it is not necessary to correct for the SSD dependence.

The Field size, specified at isocenter, will affect the dose response of the diode due to changes in scattered radiation, both from inside the treatment head and from the patient/phantom. Because of the energy dependence the diode will overestimate the contribution from the lower energy scattered radiation.

Diode sensitivity increases with temperature. The sensitivity variation with temperature is due to changes in the semiconductor band gap. Differences in calibration temperature and the temperature at measurement (such as when the diode is placed on the skin of the patient) will affect the measurement. The sensitivity increase with temperature is linear.[22]

High angle of incidence will affect the diode response. The asymmetrical design of diode detectors will result in a intrinsic directional dependence. In addition to this intrinsic dependence the change of scatter conditions might affect the response.

1.3 EqualDose

The QA software EqualDose is used for independent verification of the plans generated by the treatment planning system. It uses a multi source model, describing the contribution from different components inside the treatment head, to simulate the energy fluence leaving the treatment head. The absorbed dose inside the irradiated object is calculated through a pencil beam algorithm. The irradiated object is represented as a rectangular block of water with the surface always perpendicular to the central beam axis. EqualDose will base the model on several parameters such as collimator properties, measured output factors, TPR\(_{20,10}\) and a dose profile. For more information about the algorithms behind EqualDose, see a series of articles by Olofsson and Nyholm.[23, 24, 25, 26, 27, 28]

EqualDose has two main applications, treatment plan verification and in vivo verification. In treatment plan verification the dose to a selected point is calculated for each beam and compared to values provided by the treatment planning system. This independent check can detect errors in the treatment planning program and in the export of treatment plans.

During in vivo verification a measurement point is chosen by the user and EqualDose calculates the expected detector readout. One of the advantages with EqualDose is that the lateral fluence distribution is shown for each field, see figure 1.2. This simplifies the process of choosing a suitable measurement point during IMRT. A printout listing the measurement points can be given to the personnel performing the measurement, see figure 1.3.
1.3. EqualDose

Figure 1.2: The layout of the in vivo function in EqualDose. The 2D plot on the right side shows the fluence and the measurement point for the selected beam. On the left side the beams are selected.

**Beam: GV 210 (1)**

**In-vivo point:**
- **Invivo**
- Distance to central axis:* 40 mm
- SSD: 921.5 mm
- X**: 40 mm
- Y**: 0 mm
- Z**: 78.5 mm
- Measurement:
- EqualDose (calc): 42.5
- Rel.diff
- Detector: EDP-10
- Gantry

Figure 1.3: The in vivo printout produced by EqualDose. The picture on the right shows the measurement point and the shape of the first segment for that beam. The information on the left side includes the measurement point coordinates, and the calculated diode reading at the given SSD.
1.4 Aim of the report

The aim of this work is to study the characteristics of two diode detectors, EDP-10 and RDP by IBA dosimetry, and evaluate the clinical suitability of the two detectors. EDP-10 is currently widely used for in vivo dosimetry in an energy range between 4 and 8 MV. RDP is a recently developed detector using the same chip as the EDP series, but utilizing a cylindrical design in an attempt to create a detector independent of incident angle (see figure 2.2). Additionally, we will evaluate the feasibility of performing routine in vivo measurements in SMLC-IMRT (step-and-shoot) with the aid of the EqualDose QA-program.
Chapter 2

Materials and methods

2.1 Equipment

2.1.1 Detectors

During this work two types of detectors were used, EDP-10\textsuperscript{3G} and RDP. The 3G indicates the third generation of Scanditronix chips which has been in use since 2001. Both detectors contain the same type of chip, a p-type pre-irradiated silicon diode with approximately 50\(\mu\)m effective thickness of the ionization volume. Both detectors use a build-up cap of stainless steel. EDP-10 uses a hemispherical build-up cap while RDP uses a cylindrical design, see figure 2.1.

\begin{figure}[h]
\centering
\includegraphics[width=0.8\textwidth]{detectors.png}
\caption{Schematics of the detectors}
\end{figure}

2.1.2 Accelerator

Unless noted otherwise, the measurements were performed using a Siemens ONCOR linear accelerator, providing a 6 MV photon beam \((TPR_{20,10}=0.675)\)

2.2 Basic detector characteristics

Unless noted otherwise, the detectors were connected to an Elektra precision electrometer (Precitron AB).

2.2.1 Field size dependence

The diode was placed on a back scatter phantom at \(SSD=100\text{cm}\). A number of square fields ranging from 4\text{cm} to 30\text{cm} were measured. The field size dependence \((FSD(c))\) was calculated according to 2.1 with reference field size \((c_{\text{ref}})\) of 10x10cm\(^2\).

\[
FSD(c) = \frac{M(c)}{M(c_{\text{ref}})} \frac{1}{S_{\text{cp}}(c) TPR_{1.5,10}(c)}
\]  

Where \(M\) is the dosimeter reading, \(S_{\text{cp}}(c)\) is the in water output ratio (the ratio of the absorbed dose at the measured field size and the reference field size, at the reference depth 10\text{cm} in a water phantom), and \(TPR_{1.5,10}(c)\) us the TPR ratio between 1.5\text{cm} and 10\text{cm} depth. In other words: the signal is normalized at field size 10x10cm\(^2\), the signal variation with field size is compared to the dose variation with field size at 1.5\text{cm} depth in water. The dose variation is not measured directly but calculated from previously measured output ratios and TPR ratios.

However, the field size dependence is not directly used for in vivo dosimetry at the clinic in Umeå. The QA-program EqualDose handles the field size correction by choosing an effective depth of measurement; the depth at which the dose calculations best approximates the measured field size dependence of the diode detector.
2.2.2 SSD dependence

The diode was placed on a back scatter phantom. In order to study the effect of removing the electron contamination all measurements where repeated with a mini-phantom covering the detector. The mini-phantom was a 10cm high PMMA cylinder with a 4cm diameter. Measurements where performed for field sizes $5 \times 5 \text{cm}^2$, $10 \times 10 \text{cm}^2$ and $20 \times 20 \text{cm}^2$ (at isocenter) and for SSDs 80cm, 100cm and 120cm. The SSD dependence was defined as the deviation from the inverse-square law.

EqualDose handles the inverse square law in a more advanced way. In the multi-source model some (5-10%) of the energy fluence originates outside of the x-ray target consequently having a smaller SSD. The electron contamination and dose rate dependence is not taken into account. For comparison the measurements were also divided with the expected values calculated by EqualDose.

2.2.3 Angular dependence

The diode was placed on a cylindrical PMMA phantom with a 10cm radius. The choice of a cylindrical phantom was made in order to approximate the conditions during head and neck treatments. Both the axial and tilt angular dependence were measured for both detectors, see figure 2.2. Measurements where made from normal incident angle($0^\circ$) up to $60^\circ$ for axial angles and to $60^\circ/300^\circ$ for tilt angles.

EqualDose does not take incident angle into account.

2.2.4 Temperature dependence

The RDP was placed on a phantom consisting of a plastic container filled with water at desired temperature. A series of measurements where performed for each phantom temperature in order to study the sensitivity as the temperature of the diode increases. The measurements where performed with SSD 100cm and field size $10 \times 10 \text{cm}^2$.

When studying the EDP-10 the measurements were performed with some modifications to the method described above. During these measurements four different EDP-10 diodes were used at the same time. In order to simultaneously attach them to the phantom the diodes were attached to a thin piece of paper. The EDP-10 detectors were connected to an Apollo 5 electrometer.

Equal dose does not take temperature variations into account.

2.2.5 Off-axis dependence

The measurements were made with a water phantom 3D-scanner, RFA 300 (Scanditronix Medical AB). In-line and cross-line profiles were measured for $30 \times 30 \text{cm}^2$ and $10 \times 10 \text{cm}^2$ field sizes. Measurements were performed with a cylindrical RK ionization chamber (Scanditronix) with 4mm diameter and $120 \text{mm}^3$ volume, EDD (a "naked" RDP detector without build-up cap), an RDP and an EDP-10. The ionization chamber and the EDD were scanned at a depth of 2cm, while the RDP and EDP-10 were scanned at the surface of the phantom. The profiles were scaled to isocenter distance and normalized to the signal at the central axis.

Figure 2.2: Diode placed on a cylindrical PMMA phantom. This experimental setup was used for both the cylindrical RDP and the hemispherical EDP-10.

Figure 2.3: Orientations of the detectors and scan directions. The ion chamber is cylindrical and standing.
2.3 IMRT measurements

All IMRT measurements were made using actual head and neck treatment plans. The detectors were connected to an Apollo 5 electrometer (Precitron AB).

Using EqualDose a point of measurement was chosen for each beam. When the point of measurement was chosen, three aspects were taken into consideration. Firstly, a high intensity area is preferred as a lower signal leads to larger relative uncertainties. Secondly, the measurement point needs to be at a distance from any high gradient areas. During this work we aimed for a distance of at least 7.5mm. However, for some particularly difficult beams a point 5mm from the nearest high gradient region was chosen. Thirdly, the ease of placement needs to be considered. Complicated points of measurement can affect the precision of diode placement; especially in clinical practice when the staff has to consider time restraints. In this work I aimed to place the point of measurement along either the X-axis or the Y-axis and as close to the central axis as possible. For some complicated beams the point of measurement was placed outside both axes but close ($\leq 5\text{mm}$) to one axis.

2.3.1 IMRT using a flat phantom

Measurements using a flat phantom were performed as a first test of the concept. In this case, the ease of placement was not taken into account when choosing the points of measurements. The diode was placed at a flat back scatter phantom at SSD 90cm. The treatment plans were readjusted to SSD 90cm and $0^\circ$ gantry angle. Five plans, each containing six to seven beams, were used. Each beam was measured once with the EDP-10 and once with the RDP.

2.3.2 IMRT using a cylindrical phantom

A cylindrical PMMA phantom with a 20cm diameter was used, resulting in SSD 90cm for all angles. Four plans, each containing seven beams were used. The points of measurement was chosen according to section 2.3. Each beam was measured once with the EDP-10 and once with the RDP.

2.3.3 IMRT verification in vivo

In vivo measurements were performed on three patients. Three of four beams were measured for each patient. The beams included posterior beams (gantry angle $\approx 150^\circ$), lateral beams (gantry angle $\approx 70^\circ$ and $\approx 50^\circ$) and anterior beams (gantry angle $\approx 0^\circ$). Each beam was measured twice with the EDP-10 and twice with the RDP. Points of measurement were chosen according to section 2.3.
Chapter 3

Results

3.1 Basic detector characteristics

3.1.1 Field size dependence

Both detectors overestimated the dose for field sizes larger than 10x10cm$^2$, see figure 3.1. For most treatments field sizes are smaller than 20x20cm$^2$. EDP-10 showed an overestimation of about 1% within this range while RDP showed a better agreement with an overestimation of about 0.5%. For 5x5cm$^2$ fields EDP-10 was accurate while RDP overestimated the dose with about 1%.

Compared to values calculated by EqualDose the measured values where slightly (about 0.5%) higher for field sizes below 10x10cm$^2$, see figure 3.2. For 20x20cm$^2$ the values measured with EDP-10 was about 1.5% higher, and the values measured with RDP about 1% lower than their respected values calculated by EqualDose.
3.1.2 SSD dependence

The sensitivity variation with SSD, see figure 3.3 to 3.6, was less than ±1% for both detectors. When compared to values calculated by EqualDose, see figure 3.6, the deviation was also less than ±1%.

Figure 3.3: Relative sensitivity for field size 5x5cm²

Figure 3.4: Relative sensitivity for field size 10x10cm²

Figure 3.5: Relative sensitivity for field size 20x20cm²

Figure 3.6: Relative sensitivity divided by expected values from EqualDose.
3.1.3 Angular dependence

For axial angles the EDP signal remained constant up to 30°, see figure 3.7a. At larger angles the signal decreased gradually to about 98.5% at 60°. For tilt angles, see figure 3.8a, the signal decrease started at smaller angles and was about 97% for 60°.

The RPD showed a minimum signal of about 99.5% at 40° axial incident angle, figure 3.7b. The tilt angle measurements, see figure 3.8b, show a steep variation in signal at small angles. There was a maximum value of about 112% at 30° tilt angle and a minimum of about 97.5% at 30°.

Figure 3.7: The relative signal with axial angles.

Figure 3.8: The relative signal with tilt angles.
3.1.4 Temperature dependence

The EDP-10 measurements with the Apollo 5 electrometer can be seen in figure 3.9. Diode 2 was completely new, diodes 1, 3 and 4 had been used for a long time. The EDP-10 diodes showed a sensitivity variation with temperature between 0.1 and 0.3 \(^{\circ}\)C. All EDP-10 diodes had reached their maximum signal after 3.5 minutes; detector 4 and 2 show a slightly higher signal at 8 minutes but this increase lies within the uncertainty of the measurement.

Measurements were performed with the RDP diode for two phantom temperatures, see figure 3.10. For both temperatures the sensitivity variation with temperature was about 0.2 \(^{\circ}\)C. The maximum signal was reached after about 7 minutes.
3.1.5 Off-axis dependence

For 30x30cm$^2$ field sizes the EDP-10 overestimated the relative dose at large distances from the central axis, see figure 3.11b and 3.12b. For cross-line measurements with 30x30cm$^2$ field size, see figure 3.11b, the RDP and EDD showed asymmetric off-axis dependence, overestimating the dose on one side of the central axis and underestimating the dose on the other. For 30x30cm$^2$ in-line measurements, see figure 3.12b, the RDP and EDD agreed with the ion-chamber measurements. None of the detectors showed any notable off-axis dependence for 10x10cm$^2$ fields, see figure 3.11a and 3.12a.

Figure 3.11: Cross-line profiles

Figure 3.12: In-line profiles
3.2 IMRT measurements

3.2.1 IMRT using a flat phantom

Figure 3.13: Deviation from expected value using a flat phantom.

3.2.2 IMRT using a cylindrical phantom

Figure 3.14: Deviation from expected value using a cylindrical phantom.
3.2.3 IMRT verification in vivo

The in vivo measurements using EDP-10 diodes generally resulted in deviations within ±5%, see figure 3.15a. However, one outlying value was seen at -18.2%. During the first measurement for this field the measured value was within 0.2% from the expected value. Because of this, the outlying value was attributed to misplacement of the diode. Two standard deviations are reported for EDP-10, with and without the outlying value.

![Deviation from expected value for EDP](image)

Figure 3.15: Deviation from expected value for EDP.
Chapter 4

Discussion

4.1 Basic detector characteristics

4.1.1 Field size dependence

The overestimation of dose (about 1%) for large field sizes were not large enough to make the use of correction factors for field size essential but it would improve the accuracy for both detectors.

Similar results is shown for EDP-20 diodes by Norberg[30] who reports an overestimation slightly below 1% for 20x20cm$^2$. Nilsson[31] shows an overestimation of 1% for 5x5cm$^2$, 15x15cm$^2$ and 20x20cm$^2$ field sizes with an older model EDP-10.

4.1.2 SSD dependence

The SSD dependence measurements with the mini-phantom (figures 3.3 to 3.5) showed little difference from the measurements without mini-phantom. This indicates that signal due to electron contamination is very small for both detectors and thus not a problem at 6MV beams. By choosing a suitable calibration SSD the effect of the SSD dependence can be decreased. In head and neck treatments most beams are delivered with a SSD close to 90cm, with some beams having SSDs between 80cm and 95cm. By choosing SSD 90cm for calibrations the SSD dependence within the treatment distances will be less than ±0.5% for both detectors and no SSD correction factors are needed.

The size of the SSD deviation agrees with results obtained by Saini and Zhu[21] for EDP-10 and by Norberg[30] for the EDP-20. The most remarkable result is that all EDP-10 measurements show lower sensitivities at shorter SSDs. All mentions in the literature show an increase in sensitivity for shorter SSDs. The error in my result corresponded to a 4mm error in SSD. It is unclear what caused this sensitivity decrease for shorter SSDs. From the literature and from the superlinearity of silicon diodes we suspect that our values are incorrect.

4.1.3 Angular dependence

During the in vivo measurements large angles of incidence were sometimes encountered. For a majority of the fields the angle of incidence was estimated to be less than 40°, an exception was the frontal fields sometimes having close to 60° angle of incidence.

While the RDP had almost no signal variation with axial angles the signal variation with tilt angles was considerable. For axial angles of only ±10° the sensitivity varied between about 97.5% and 102%, which was larger than the deviation for EDP-10 at 60°. When placing the detector on a patient it will never be completely horizontal, which means that angles of ±10° must be expected. Under these conditions the RDP will have a larger angular dependence than EDP-10, even though it had a very low axial angular dependence.

In figure 3.7a a sensitivity slightly above 100% is seen for 10° and 20°. However, this small increase is within the error of measurement. From this result it seems that EDP-10 have no noticeable dependence on axial angles below 30°. For tilt angles the dependence is slightly larger and the sensitivity drop starts earlier, for angles around 20°. During most in vivo measurements the angle of incidence will be less than 40°, resulting in an angular dependence of less than 1%. For some in vivo measurements the angle of incidence will be close to 60°, resulting in an angular dependence somewhere between 1-2%. Due to difficulties in determining the angle of incidence, no correction factors is used. An angular dependence of more than 1% is not ideal but will still give an acceptable accuracy.

4.1.4 Temperature dependence

While studying the temperature dependence we encountered both some expected and some rather strange results. As seen in figure 3.10 the RDP showed a 0.2% SVWT, which agrees with the
EDP-10 sensitivity variation reported by Rikner and Grusell[18]. One difference between RDP and EDP-10 can clearly be seen. While they have the same SVWT the RDP reached thermal equilibrium after 7 minutes, compared to the 3.5 minutes it takes for EDP[17]. This is probably because the cylindrical design results in a smaller contact area.

The EDP-10 measurements proved to be problematic. When measured with the Apollo 5 electrometer, see figure 3.9, the EDP-10 behaved as expected from the literature, i.e. the SVWT were between 0.1 and 0.3 °C. The measurements with the Elektra electrometer, however, showed a large negative SVWT, -0.6 to -0.8 °C. The Elektra measurements where performed with two different EDP-10 diodes and was repeated two weeks after the first measurement. The two diodes that showed a negative SVWT with the Elektra electrometer showed a normal positive SVWT with the Apollo 5 electrometer. Because of this the error seems to be due to the electrometer or the cables. However, we have not been able to explain why the RDP did not have the same problem.

4.1.5 Off-axis dependence

As seen in figure 3.11a and 3.12a, the off-axis placement does not have a notable influence on the sensitivity for 10x10cm² fields. For very large fields, see figure 3.11b and 3.12b, EDP-10 shows no notable off-axis dependence if placed within 5cm from the central axis.

The effects of the cylindrical design of the RDP and the resulting angular dependence can clearly be seen for 30x30cm² fields. When the detector was moved perpendicular to it’s axis, see figure 3.12b it showed a very good agreement with the ion chamber, even close to the field borders. On the other hand, when the detector was moved along it’s axis, see figure 3.11b, it showed a large asymmetric off-axis dependence. The same behavior was seen for a cylindrical detector without the build-up cap (EDD).

4.1.6 Improvements on the RDP design

The RDP detector was shown (see section 4.1.3) to have larger angular dependence than EDP-10 under standard treatment conditions. Since the RDP was designed to be independent of incident angle, the design ought to be improved, addressing the high dependence on tilt angles. One way is to use a higher density absorber material to shield the chip, see figure 4.1. Another option is to use a dual chip design, with two diodes facing different directions, creating a more symmetric design.

4.2 IMRT measurements

The phantom measurements, see figures 3.13a and 3.13b, showed an overestimation of about 1.5-2%. This might in part be explained by EqualDose underestimating the signal for field sizes below 10x10cm². The precision was good; all measurements were within 5 percentage points of the median value. One of the most notable problems encountered when performing in vivo measurements for IMRT is the diode placement. Both the choice of measurement point and accurate placement on the patient are crucial. The large deviation shown in one EDP-10 measurement (see figure 3.15a) was encountered when the point of measurement was placed within 5mm from the dose gradient seen in EqualDose. The -18% deviation corresponded to a 5mm misplacement of the point of measurement. However, it is important to remember that EqualDose calculates the detector signal in a point, while the silicon chip has a diameter...
of about 2.5mm and the EDP build-up cap diameter of 10mm. This means that in practice the misplacement of the diode does not have to be as large as 5mm to result in the same deviation. Only one of the measurements showed this large deviation, the other EDP-10 measurement for the same field agreed with the expected value. This illustrates the need for great care when placing the diode and the personnel will have to be aware of that importance. It also shows that the point of measurement should be more than 5mm away from the closest significant dose gradient. The cylindrical design of the RDP diode might prove useful when placing the measurement point closer to gradients. Since the chip is placed on end the very low effective thickness of the ionization volume means that the detector can be seen as having no extent in one direction.

This trade-off between ease of diode placement on one hand and choosing a good measurement point (with high enough fluence and away from gradients) on the other hand is sometimes hard to balance. For most fields the measurement point was placed along one of the two main axes. Because of the high risk of placement errors when placing the diode away from both axes this should be avoided if possible. For some fields this was not considered possible. In these cases the point of measurement was placed close (preferably 5mm or 10mm) from one of the axes. It is also worth noting that the settings in the treatment planning software will affect the chance of finding a good point of measurements. If many small segments are allowed it may make the choice of measurement point very problematic. At the clinic in Umeå the minimum segment size allowed is 5cm².

The in vivo measurements showed good agreement with values calculated by EqualDose. With EDP-10 90% of the measurements were within 5% of the calculated value, with RDP this was true for 85% of the measurements. This precision is about the same as for conventional treatments.[32]

There are few studies of diode detectors used for in-vivo dosimetry during IMRT. In one study Alaei et al. report that for head and neck treatments, 22% of the measurements were within 5% of the expected value and 72% of the measurements were within 10±%.33 They used a method where the diode is placed on the central axis and the dose maximum along the central axis is taken directly from the treatment plan. Based on their results, the use of EqualDose and the method described in this thesis will greatly improve precision. The drawback of using the EqualDose method is that diode placement requires more effort. However, the increase of precision should justify this additional effort.
Chapter 5

Conclusions

The results indicate that it is possible to perform in vivo dosimetry for SMLC-IMRT using the quality assurance software EqualDose. Measurements with both the EDP-10 and the RDP diodes showed good enough accuracy to enable a 5% tolerance limit.

While being essentially independent of axial angles of incidence, the RDP showed a very high dependence on tilt angles. In practice the RDP will have higher angular dependence than the EDP-10. An advantage with RDP is that the orientation of the chip can allow measurement points closer to significant dose gradients.

5.1 Limitations and Future work

This work has only investigated treatments with 6MV photon beams. Measurements at higher energies (15-20MV) might prove more problematic due to a large increase of the electron contamination. It would be interesting with another study, containing more patients and both low and high energy beams.

It would also be interesting to evaluate the feasibility of performing in-vivo dosimetry for DMLC-IMRT. In DMLC-IMRT the dose gradients are even more problematic so it is doubtful if diode dosimetry will work. EqualDose supports dynamic techniques but only SMLC-IMRT is used at the clinic in Umeå.
Chapter 6

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References


[29] *EQUAL-Dose Manual*.


