Dose distribution beneath the Leipzig skin applicator set

Thesis for Master of Science in Medical Radiation Physics

Sebastian Sarudis
Abstract

The Leipzig skin applicator set is used for treatment of small superficial lesions (with diameters < 25 mm) together with an $^{192}$Ir HDR source. The dose calculation formalism suggested by the AAPM TG43 group for brachytherapy sources was used to calculate the dose rate profiles for different applicators at depths ranging from 0 - 0.5 cm beneath the skin surface. The obtained profiles were compared with Monte Carlo profiles and with actual measurements obtained with radiochromic film in order to assess the usefulness of the TG43 model for the special case of the Leipzig applicators. Furthermore, a modification of the TG43 model (TG43-EQ) was made in order to take some of the special characteristics of the Leipzig applicators into account and to improve the agreement with the Monte Carlo model. The results showed a mean difference between the TG43 and the MC of 1.25 % for the horizontal applicators and 2.11 % for the vertical applicators decreasing to 1.22 % for the horizontal and 1.75 % for the vertical applicators for the TG43-EQ model.

Alarming differences between the theoretical distributions (assuming perfect set-up) and the experimental results were discovered. A tilt in the vertical source position was shown to decrease the total Biological Effective Dose to the tumour with up to 9 % for a regular treatment with the Leipzig applicator, 6 fractions with 6 Gy per fraction. This difference might be even larger for real clinical treatments. A displacement in the SSD for the horizontal applicators was also measured leading to overdosage of the tumour.
Acknowledgements

First of all I would like to thank my principal supervisor Dr Leonard Wee for all his time and patience and for the help, support and valuable tips throughout the entire project.

I would also like to thank my co-supervisor Dr Sean Geoghegan and the other physicists at the department, Mr Simon Woodings, Mr Peter Rampant and Dr Adrian Perry for their support and encouragement during my time here.

A special thanks goes to Dr David Causer without whom I would not have had the opportunity to come here in the first place.

Last but not least I also want to thank the rest of the staff at the department of Medical Physics here at the Royal Perth Hospital for their kindness and for making this a memorable experience. Thanks to you I will always think of how kind and how helpful Australian people are when I think back on Australia.
1. Introduction

Radiation therapy for superficial skin lesions has been in use for many years and has demonstrated local control with excellent cosmetic outcome.

The preferred modality is soft x-rays (30 - 120 kVp) generated by a superficial unit. However in recent times, alternatives using radioactive sources in close contact to the skin have been developed as an alternative to the superficial x-ray treatments. The advantages of using radioactive sources for superficial skin lesions are a more flexible treatment unit, since the applicators used during treatment are small and easy to handle and fast dose fall-offs due to the short distance between the source and the target. The disadvantage is that only small areas can be treated due to the fast dose fall-off. Treating large areas with a short SSD would lead to unacceptable high doses to the areas close to the source in order to obtain a sufficient high dose to the areas far away.

In contrast to treatments with radioactive sources, superficial units can be used to treat larger areas and lesions at larger depths. However, as superficial x-ray units are not fully used in many clinics, the use of the current applicators can make the superficial units redundant since the applicators are used together with brachytherapy HDR units, which are commonly available and used for other types of treatments as well.

Another alternative for treating superficial lesions is the use of electron beams produced in a linear accelerator. However, small superficial malignancies are in most cases difficult to treat with electron beams since the shallow depths and small sizes of the tumours makes it necessary to use bolus to modify the depth dose in order to deliver maximum dose to the lesion. Collimation using electron cut-outs and extra shielding on parts of the face is sometimes needed which complicates the treatment procedures. Furthermore, the linear accelerators are in high demand for other types of treatments.

There are different kinds of surface applicators that can be used in combination with a radioactive source in the proximity of the treatment area. One model suggested by Kron et al. 2002, is the use of a cylindrically shaped surface lead applicator together with a circular shaped flattening filter. The applicator is used together with a small cylindrically shaped $^{192}$Ir source that is driven into the desired position by a HDR
brachytherapy unit. The use of radioactive sources close to the treatment area is suitable for superficial lesions since the short distance and low energies of the photons results in fast dose fall-offs which helps protect underlying healthy structures from being irradiated (Evans et al. 1997, Niu et al. 2004, Perez et al. 2005, Kron et al. 2002). However, the close distance of the source also causes considerable dose inhomogeneities beneath the applicators. Kron et al. 2002, suggests using thin sheets of lead foils to shape the dose distribution beneath the applicators in order to get a more uniform dose distribution. The dose distribution obtained with the lead foil filters is within ± 5% of the dose at the central axis for a circular field size with a diameter of 30 mm. By using Monte Carlo calculations the filters can be optimised since the multiple photon energies of $^{192}$Ir, the scatter from the lead collimator and the actual shape of the source can be taken into account. This has been done by Jeraj et al. 2002, and the uniformity of the dose distribution achieved was improved to only ± 1 % difference from the dose at central axis. However, the filter is used at the expense of a very much lower dose rate, hence longer treatment times.

This type of skin applicator also has the advantage that the source to surface distance (SSD) can be changed in order to model the dose distribution since there are three different positions in which the source catheter can be inserted.

Another type of skin applicator that is available for clinical treatment is the Nucletron Leipzig applicator set. The Leipzig applicators were first mentioned in 1995 and have been a subject for research ever since (Evans et al. 1997, Niu et al. 2004, Perez et al.
The Leipzig applicator set consists of six different cone-shaped applicators (Perez et al. 2005) with different sizes and dosimetric characteristics. Each applicator is constructed from a tungsten/steel alloy and consists of the cone-shaped collimator and a plastic protective cap that absorbs the secondary electrons that are emitted from the collimator walls as primary radiation interacts. A radioactive source (\(^{192}\)Ir) is positioned close to the cone's focal spot through a catheter with the help of a brachytherapy HDR remote afterloader unit. The longitudinal axis of the radioactive source can be positioned either parallel or perpendicular to the treatment surface depending on the applicator type used and the cone exit diameter can be chosen to be 10, 20 or 30 mm depending on the size of the treatment area. Combining these two characteristics, cone exit diameter and source orientation relative to the treatment surface, each skin cone has its own characteristic dose distribution.

To be able to treat malignancies effectively it is important to accurately know the dose distribution beneath each of these applicators. Calculation of the dose distribution is not trivial because of the complex geometry of the Leipzig applicators and the multiple photon energies of \(^{192}\)Ir. The emitted photons interact with photoelectric effect, coherent scattering and with Compton interactions in the collimator walls and in the source encapsulation material, and produce secondary photons and electrons that will contribute to the dose. Furthermore, the cylindrical shape of the \(^{192}\)Ir source and the thin steel capsule within which the source is encapsulated has to be accounted for. All this has to be taken into consideration if the dose calculations are to be accurate.

The dosimetric characteristics for each of the applicators have previously been investigated by comparing Monte Carlo simulations to experimental measurements with radiochromic film (Niu et al. 2004). Radiochromic film is a type of film which is nearly tissue equivalent (above 100keV), has a high spatial resolution and its dose response appears to be energy independent (Zhu et al. 1997, Klassen et al. 1997, Schumer et al. 1999, www.ispcorp.com, Chiu-Tsao et al. 2004, Kron et al. 2002). The measurements in this study were done using this type of film since it has previously been used for similar types of measurement and the film's characteristics have been thoroughly investigated (Zhu et al. 1997, Klassen et al. 1997, Schumer et al. 1999, www.ispcorp.com, Chiu_Tsao et al. 2004, Suchowerska et al. 1999, Kron et al. 2002).

Besides measuring the dose profiles beneath the six available Leipzig applicators a comparison between the measured absorbed dose and a dose calculation formalism for
brachytherapy sources suggested by the American Association of Physicist in Medicine (AAPM) (Nath et al. 1995, Rivard et al. 2004) has been made. A quantitative agreement between the measured results and the dose calculation formalism suggested by the AAPM would give a much quicker way to estimate the dose rate at different points beneath the Leipzig applicators instead of using Monte Carlo simulations. This quick check could then be used in clinical situations if the dose plan for a patient suddenly has to be changed due to, for example, misplacement of the applicators or just for quick checks to ensure that the dose to the lesion is sufficient.

The aim of this project is to characterise the dose distribution beneath each of the six skin applicators in order to determine if the AAPM dose calculation formalism for brachytherapy sources can be appropriately applied in this particular situation, and to investigate the spatial dose differences between idealised set-ups and real results obtained during clinical use.

1.1 The Leipzig surface applicators

The Leipzig skin applicators were designed for use with the Nucletron microSelectron-HDR remote afterloader system as an alternative method to treat superficial malignancies and lesions. The applicator set consists of six different cone-shaped applicators (Perez et al. 2005). The choice of applicator depends on the size of the lesion and the desired dose distribution. Besides the simplicity of using the Leipzig applicators, the advantages, when used for superficial lesions, over superficial/orthovoltage x-ray treatments are sharp dose gradients at the edges of the treatment area due to the tungsten/steel alloy collimators and faster dose fall off due to the short SSD that is used (1.6 cm for the Leipzig applicator compared to 40 cm for superficial units). This leads to lower doses to the healthy tissue and structures surrounding the treatment area. The disadvantages are that the applicators are only suitable for small treatment areas since the largest cone diameter available is only 30 mm, the dose distribution is inhomogeneous since no flattening filter is used and that the use of the applicators is limited to only one treatment SSD.
The Leipzig applicator consists of a tungsten/steel alloy shaped cone with an $^{192}\text{Ir}$ source located at the focal spot of the cone (1.6 cm above the skin surface), as shown schematically in Fig 1.2. The source orientation varies depending on the type of applicator, with the longitudinal axis of the source aligned either parallel or perpendicular to the treatment surface. There are three different horizontal applicators with the cone diameter of 10, 20 and 30 mm (H10, H20 and H30) and three vertical applicators with the same diameters as the horizontals (V10, V20 and V30). Figure 1.3 shows a photography of the H30 and V30 applicators. Each applicator has a 1 mm thick protective plastic cap that should always be used during treatment since it absorbs the secondary electrons produced in the collimator walls. This is very important since the surface dose can increase by a factor of ten if the protective cap is removed (Evans et al 1997) leading to unnecessary high doses to the skin.

![Figure 1.2](image)

*Figure 1.2: Schematic figure of the Leipzig applicators with the dimensions in mm. The left hand figure shows the horizontal applicator type with the source entering parallel with the treatment surface and the right hand figure shows the vertical applicator type with the incoming source perpendicular to the treatment surface.*
The source model that is used together with the Leipzig applicators is the so-called v2 source model. This source model comes together with the microSelectron-HDR remote afterloader unit (Daskalov et al. 1998) and the dimensions of the source and the encapsulating material are shown in Fig 1.4. The radioactive $^{192}$Ir is uniformly distributed in the source core and the capsule surrounding the source is an AISI 316L steel capsule.

![AISI 316L Steel Capsule](image)

Figure 1.4: The schematic design of the $^{192}$Ir v2 source model used with the microSelectron HDR remote afterloader. The Iridium is uniformly distributed in the source core. The figure is taken from Daskalov et al.1998.
1.2 Thesis Outline

The aim of this thesis is to compare the absorbed dose to water predicted by the TG43 dose calculation formalism to Monte Carlo simulated profiles and real measured results for the Leipzig skin cone applicator set in order to decide the usefulness of the TG43 model in the special case of the Leipzig skin cone dosimetry. Furthermore, the dose differences between the idealised set-ups and real results obtained during clinical use are examined to determine whether the assumed dose distributions are correct for real clinical situations. The measured doses have been obtained using two different types of radiochromic film, GafChromic MD55 and GafChromic HS film, since they were found to have the appropriate characteristics.
2. Preparation of radiochromic film and scanner

2.1 Radiochromic film

Radiochromic film is a new type of film used in radiotherapy dosimetry. It is a colourless film with a nearly tissue equivalent composition and high spatial resolution (Zhu et al. 1997, Schumer et al. 1999, www.ispcorp.com), which makes it very useful for measuring the absorbed dose in steep dose gradients, such as those found in brachytherapy (Klassen et al. 1997, www.ispcorp.com). The most common type that is currently used is the GafChromic film. The GafChromic film is manufactured by ISP Technologies Inc. and exists in various formats for different purposes (www.ispcorp.com). The advantages of GafChromic film are that its response to dose is fractionation and dose rate independent (Schumer et al. 1999, www.ispcorp.com), and it is also relatively independent of energy above 100 keV (Klassen et al. 1997, Schumer et al. 1999, www.ispcorp.com). It has high spatial resolution and it is ideal for measuring two dimensional dose distributions in steep dose gradients (Klassen et al. 1997, Schumer et al. 1999, www.ispcorp.com, Zhu et al. 1997). Furthermore it is relatively insensitive to normal room light. Since it is self-developing the need for dark room or film processor is eliminated (Schumer et al. 1999, www.ispcorp.com). The disadvantage is that the uncertainties in the measurements can be as high as 8% due to a non-uniform distribution of the radiation sensitive monomer within the same film sheet. The non-uniform distribution of the active material of the film can therefore lead to a non-uniform dose response for a GafChromic film sheet (www.ispcorp.com). Therefore careful calibration of the film and scanner is necessary before usage.

The active component of the GafChromic films is a microcrystalline, radiation sensitive monomer that reacts and produces a dye polymer upon exposure. The dye polymer gives the film the characteristic cyan blue colour it develops when exposed to radiation. The amount of dye polymer produced, and thus the opacity of the film, is proportional to the absorbed dose to the film (www.ispcorp.com). The polymer production mainly occurs when the active component is exposed to high energy photons, thus the film is not particularly sensitive to visible light. Numerous studies have been made on the normal room light effect on GafChromic film and it has been found that it has a small effect on the film darkening. If the film is exposed to
normal room light with the mean intensity of 1000 lux, the density changes by approximately 0.00086 density units for the MD55 model and 0.0056 density units for the HS model per 24 hours (www.ispcorp.com). This change is so small that the film can be left in fluorescent light for several hours without any noticeable effects. However, the film should not be left in room light for long periods since that can darken the film leading to errors when the absorbed dose is measured. Therefore it is recommended to avoid unnecessary exposure of the film to fluorescent lights.

The thickness of the sensitive part of the film varies between different models and is sandwiched between two layers of clear polyester. The structures of the GafChromic MD55 and HS film models are shown in figures 2.1 and 2.2. These are the two formats of radiochromic film that are used in this thesis.

<table>
<thead>
<tr>
<th>Clear Polyester - 66 μm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Active Layer 16 μm</td>
</tr>
<tr>
<td>Adhesive - 25.4 μm</td>
</tr>
<tr>
<td>Clear Polyester - 25.4 μm</td>
</tr>
<tr>
<td>Adhesive - 25.4 μm</td>
</tr>
<tr>
<td>Active Layer 16 μm</td>
</tr>
<tr>
<td>Clear Polyester - 66 μm</td>
</tr>
</tbody>
</table>

Figure 2.1: The composition and structure of the GafChromic MD55 film model

<table>
<thead>
<tr>
<th>Clear Polyester 96.52 μm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Active Layer 40 μm</td>
</tr>
<tr>
<td>Clear Polyester 96.52 μm</td>
</tr>
</tbody>
</table>

Figure 2.2: The composition and structure of the GafChromic HS film model
As can be seen from the figures above the structure of the two film models is quite different. In the MD55 model the sensitive layer is divided into two pieces separated by a thin layer of clear polyester and two thin layers of adhesive tape, while for the HS model the sensitive layer is thicker and without any layers dividing it.

Since the outer polyester layers protect the active layer of the film it can be used to measure directly in water. Water does however slowly diffuse in between the layers since they are not sealed at the edges. The rate at which the water enters the film is very slow and the film can be kept under water for up to an hour without any noticeable effects (www.ispcorp.com). If water still penetrates the film and reaches the active layer this is visible as an opaque white border close to the exposed edges and can be avoided by cutting the film piece into a slightly larger size than needed for the measurement.

To determine the absorbed dose to the film the absorption characteristics of the dye polymer is exploited. The absorption spectrum for the dye has a major absorption peak at approximately 675 nm and a minor at 615 nm and the absorption of the emitted light through the film increases as the absorbed dose to the film increases (figure 2.3).

![Figure 2.3: The absorption spectra for GafChromic MD55 and HS film models. The major and minor absorbance peaks are clearly visible at 675 and 615 nm. The figure is taken from Reinstein et al. 1997.](image-url)

These wavelengths correspond to the red portion of the visible light spectrum. Hence, the highest sensitivity to absorbed dose is achieved by measuring with red light with the wavelength close to that of the major absorption peak. Depending on the wavelength of
the light used, the relation between the Net Optical Density (NOD, chapter 2.2) and the absorbed dose is different (Reinstein et al. 1997, www.ispcorp.com). Since it is the Net Optical Density that is measured and thereafter converted to dose it is crucial to know the relation between them. This relation can however differ slightly between different film sheets and therefore it has to be established for every sheet that is used for measurements.

After irradiation the concentration of dye within the active layer increases, which in turn increases the optical density of the film. This occurs gradually over time but stabilises to an almost constant value after approximately 48 hours. This property of the GafChromic film has previously been investigated in several studies (Klassen et al. 1999, Reinstein et al. 1998, www.ispcorp.com) and one needs to keep this increase of optical density with time after exposure in mind when measuring the absorbed dose to the films. From the studies that have been made, the manufacturer recommends a waiting period of at least 24 but preferably 48 hours before reading out the optical density since this will give the film enough time to stabilise. If the measurements are not done at approximately the same time after exposure, the measured values should be corrected to take the post exposure increase in the optical density into account. In figures 2.4 and 2.5 the post exposure density change is shown for the two film models. The rapid increase during the first hours after exposure and the slower increase after 24 hours is clearly visible. During the first 24 hours after exposure the density increases with nearly 15% while it only increases with approximately 2% during the following 96 hours (www.ispcorp.com). The figures have been supplied by the film manufacturer.
One of the key features of the GafChromic film that makes it ideal for dose measurements is the energy independence. The response of GafChromic film does not depend on the energy of the photons or electrons giving the dose. This is crucial since corrections would have been necessary otherwise to correct for the energy spectra of $^{192}$Ir. This also means that the calibration curves (chapter 2.5) for each sheet can be obtained with energies other than $^{192}$Ir that is used to expose the films (www.ispcorp.com).

The main difference between the MD55 model and the HS model is the sensitivity, which is defined as Optical Density per unit of dose. As the name indicates the HS (High Sensitivity) model is specially designed to be more sensitive than the MD55 model. From the manufacturers tests the high sensitivity model has been shown to be approximately twice as sensitive as the MD55 model which makes the HS model more suitable for measuring low doses. However, we have found the MD55 more robust in handling, since HS film is prone to delamination.
A summarisation of the characteristics and properties for the two film models is found in the table 2.1 and 2.2.

<table>
<thead>
<tr>
<th>Property</th>
<th>GafChromic® MD55 Radiochromic Dosimetry Film</th>
</tr>
</thead>
<tbody>
<tr>
<td>Configuration</td>
<td>Two active layers on polyester substrates laminated with adhesive tape</td>
</tr>
<tr>
<td>Size</td>
<td>5&quot;x5&quot; minimum, other sizes upon request</td>
</tr>
<tr>
<td>Substrates</td>
<td>66 μm clear transparent polyester</td>
</tr>
<tr>
<td>Active layer thickness</td>
<td>Nominally two 16 μm thick layers</td>
</tr>
<tr>
<td>Laminating tape</td>
<td>25.4 μm polyester with double sided adhesive layers</td>
</tr>
<tr>
<td>Sensitometric response</td>
<td>Net density of 0.90 at 25Gy and 1.75 at 50Gy</td>
</tr>
<tr>
<td>Energy dependency</td>
<td>&lt;5% difference in net density for 50Gy exposures at 1MeV and 18MeV</td>
</tr>
<tr>
<td>Dose fractionation response</td>
<td>&lt;5% difference in net density for a single 40Gy dose and five cumulative 8Gy doses at 30 min intervals.</td>
</tr>
<tr>
<td>Dose rate response</td>
<td>&lt;5% difference in net density for 10Gy exposures at rates of 3.4Gy/min and 0.034Gy/min.</td>
</tr>
<tr>
<td>Stability in light</td>
<td>&lt;0.005 change in density per 1000 lux-day</td>
</tr>
<tr>
<td>Stability in dark (pre-exposure)</td>
<td>&lt;0.5x10^{-3} density change/day at 23°C</td>
</tr>
<tr>
<td>Uniformity, single sheet</td>
<td>&lt;8% sensitometric response difference</td>
</tr>
<tr>
<td>Sheet-to-sheet uniformity</td>
<td>&lt;5% difference in sensitometric response from mean</td>
</tr>
<tr>
<td>Post exposure density growth</td>
<td>&lt;12% from 1 h to 1 day after exposure, &lt;4% 1 day to 4 days after exposure</td>
</tr>
</tbody>
</table>

Table 2.1: Properties of the GafChromic MD55 film specified by the manufacturer. The values are taken from the manufacturer’s home page.
Table 2.2: Properties of the GafChromic HS film specified by the manufacturer.
The values are taken from the manufacturer’s home page

<table>
<thead>
<tr>
<th>Property</th>
<th>GafChromic® HS Radiochromic Dosimetry Film</th>
</tr>
</thead>
<tbody>
<tr>
<td>Configuration</td>
<td>Active layer between polyester substrates</td>
</tr>
<tr>
<td>Size</td>
<td>5”x5” minimum, other sizes upon request</td>
</tr>
<tr>
<td>Substrates</td>
<td>96.52 μm clear transparent polyester</td>
</tr>
<tr>
<td>Active layer thickness</td>
<td>Nominally 40 μm¹</td>
</tr>
<tr>
<td>Sensitometric response</td>
<td>Net density² of 0.80 at 10Gy and 2.15 at 30Gy</td>
</tr>
<tr>
<td>Energy dependency</td>
<td>&lt;5% difference in net density² for 20Gy exposures at 1MeV and 18MeV</td>
</tr>
<tr>
<td>Dose fractionation response</td>
<td>&lt;5% difference in net density² for a single 30Gy dose and five cumulative 6Gy doses at 30 min intervals.</td>
</tr>
<tr>
<td>Dose rate response</td>
<td>&lt;5% difference in net density² for 20Gy exposures at rates of 3.4Gy/min and 0.034Gy/min.</td>
</tr>
<tr>
<td>Stability in light</td>
<td>&lt;0.01 change in density per 1000 lux-day²³</td>
</tr>
<tr>
<td>Stability in dark (pre-exposure)</td>
<td>&lt;0.5x10⁻³ density change/day at 23°C</td>
</tr>
<tr>
<td></td>
<td>&lt;0.2x10⁻³ density change/day under refrigeration</td>
</tr>
<tr>
<td>Uniformity, single sheet</td>
<td>&lt; 6% sensitometric response difference</td>
</tr>
<tr>
<td>Sheet-to-sheet uniformity</td>
<td>&lt; 5% difference in sensitometric response from mean</td>
</tr>
<tr>
<td>Post exposure density growth</td>
<td>&lt;12% from 1 hr to 1 day after exposure, &lt;2% 1 day to 4 days after exposure</td>
</tr>
</tbody>
</table>

1. Actual thickness may vary slightly from batch-to-batch in order to match sensitivity specification.
3. Cool white fluorescent light.

2.2 Net Optical Density

A common method used to obtain the absorbed dose to a piece of GafChromic film is to measure the transmitted light intensity through it. The transmitted intensity through the film is measured and converted to Net Optical Density, NOD. Thereafter a relation between the NOD and the absorbed dose has to be established. This method has previously been used in many other studies with good results and can therefore be regarded as a reliable method (Chiu-Tsao et al. 2004, Devic et al. 2005, Kellerman et al. 1998, Klassen et al. 1999, Reinstein et al. 1997, Reinstein et al. 1998, Zhu et al. 1997).

The Optical Density through a piece of film is the 10-logarithm of the ratio between the intensity of the incident light, I₀, and the intensity of the transmitted light through the film, I.

\[ \text{OD} = \log_{10}(I₀/I) \] (2.1)
The Net Optical Density for a certain piece A, is then given by subtracting the optical density for an unexposed film piece from the optical density of piece A.

\[
\text{NOD}_A = \log_{10}(I_0/I_A) - \log_{10}(I_0/I_{\text{unexposed}}) = \log_{10}(I_{\text{unexposed}}/I_A) \tag{2.2}
\]

where \( I_0 \) is the intensity of the incident light beam, \( I_A \) is the intensity of the transmitted light through film piece A and \( I_{\text{unexposed}} \) is the intensity transmitted through an unexposed film piece.

A scanner measures the pixel value for a certain point. This measured pixel value is dependent on the light that is transmitted through the film. In order to be able to use the pixel values for calculation of the absorbed dose to the film, the registered pixel values have to be proportional to the intensity of the transmitted light ie. \( I = PV \cdot c \), where PV is the pixel value and c is a constant. If the pixel values are proportional to the light intensity through the film, the proportionality constants will cancel out when calculating the NOD and thus giving the same value of the NOD whether the intensity or the pixel value is used.

\[
NOD = \log_{10}\left(\frac{I_0}{I}\right) = \log_{10}\left(\frac{PV_0 \cdot c}{PV \cdot c}\right) = \log_{10}\left(\frac{PV_0}{PV}\right)
\]

To ensure that the transmitted light intensity is proportional to the measured pixel value, a scanner linearity test is performed before using the scanner for actual measurements.

### 2.3 Tests of scanner response

The optical density of exposed GafChromic film can be measured using a variety of methods. Various types of transmission densitometers, film scanners and photo spectrometers can be used. Each one of these items has its own sensitivity due to the light spectrum used for measurements. A densitometer using red light for measuring the optical density will result in a higher sensitivity when used together with GafChromic film than if white light is used due to the high absorption peak located in the red part of the spectrum for GafChromic film (Chiu-Tsao et al. 2004, Kellerman et al. 1998,
Reinstein et al. 1997, Schumer et al. 1999. www.ispcorp.com). These items are specially designed for film work. In recent times, conventional flat-bed scanners have also been used (Devic et al. 2005). This method is convenient since film scanners are easily available, which eliminates the need for expensive special equipment.

The use of conventional scanners to read out the optical density has previously been investigated and used in other studies with good results (Devic et al. 2005, Stevens et al. 1996, www.ispcorp.com). Reports from earlier studies have shown that the read out of the optical density of GafChromic films depends on the amount of the polarised light in the densitometer or scanner used. This means that the transmission through the film will vary when the film is rotated if the transmission light is polarised (Klassen et al. 1999). This will cause problems if the film pieces are not always positioned in the same direction. By using a film scanner this problem can be eliminated since the transmission light used is a non-coherent broadband (white) light. This makes the use of scanners both simpler and more reproducible than using equipment with linearly or circularly polarised light.

The scanner used in this study is a Hewlett Packard ScanJet II cx/T. This device has a fluorescent light panel on the cover, so light transmission through the film can be measured. The scanner is used together with a Comdek computer equipped with an Intel Pentium II processor and Windows 98 SE operating system.

The light transmission through the film is measured and since the GafChromic film has its major absorption peak in the red part of the spectrum, the obtained image should be split into the red, blue and green colour channels. Only the red channel image should be used for evaluation to obtain the highest sensitivity (Devic et al. 2005, www.ispcorp.com) for the range of absorbed doses utilised in this thesis.

Before using the scanner for actual measurements a number of checks should be done to ensure that the scanner performance is satisfactory. The scanner linearity is checked to ensure that the pixel value is proportional to the transmitted intensity, thus ensuring that a certain increase in NOD is accurately represented by the scanner. The scanner position dependence, the total scanner-bed uniformity and the dependence on the acquisition window geometry are also checked since a non-uniform or a position dependent scanner response has to be corrected for if errors are to be avoided.
2.3.1 Linearity of signal

It is important that the measured pixel value with the scanner is proportional to the intensity of the transmitted light. Otherwise a certain increase in the NOD for a film piece would not correspond to the same increase in the calculated NOD obtained from the measured scanner values, leading to errors in the calculated absorbed dose. To check this and to determine in which interval this holds true, a number of pieces were exposed to different doses in the interval between 0 - 88 Gy and the NOD was measured with a Nuclear Associates Deluxe Clamshell densitometer. By plotting the measured NOD with the Clamshell densitometer as a function of calculated NOD from the scanner, a suitable interval can be chosen in which the relation is close to linear (Fig 2.7).

![Graph](image)

Figure 2.7: The measured Net Optical Density as a function of calculated NOD for MD-55 film pieces. The film pieces were irradiated with 6 MV photons produced in a linear accelerator resulting in absorbed doses between 0 - 88 Gy. The figure shows that the relation between measured NOD and calculated NOD is only linear in the interval 0-16 Gy with a correlation coefficient of 0.99956.
The results from figure 2.7 show that the relation between the measured NOD with the Nuclear Associates Deluxe Clamshell densitometer and the NOD calculated from the scanned film pieces is different depending on the absorbed dose to the film pieces. For low doses up to 16 Gy, the relation is approximately linear, with a correlation coefficient of 0.999, meaning that the measured pixel value with the scanner is proportional to the transmitted light intensity. For higher doses the curve flattens out and the relation is no longer linear. The curve reaches saturation for doses above 56 Gy. All the film pieces used in this study are given doses below 16 Gy since it is crucial that the NOD calculated from the scanner measurements increases at the same rate as the real measured NOD if errors in the calculated absorbed dose are to be avoided.

### 2.3.2 Position dependence and whole-area uniformity

In order to have an accurate determination of the pixel value and thus the absorbed dose by the GafChromic film, the errors due to the placement of the film on the scanner bed have to be investigated. A good scanner would give approximately the same reading throughout the entire scanner bed. To check for this, an unexposed strip of film (8 x 80 mm) was placed on different parts of the scanner bed and the transmitted light intensity was measured. Eight different measurements were performed with the film strip placed in different areas of the scanner bed and the mean pixel values for the entire film strip (Fig 2.8) except for the edges, and the standard deviations were calculated for each position of the film strip. The results are presented in table 2.3. The edges of the film were left out since cutting the film into the desired shapes can affect the film's dose response properties close to the edges (Klassen et al. 1999).

---

**Figure 2.8:** Photo of the unexposed film strip used to check the scanner's position dependence. The black borders marks the area used for evaluation. The edges are left out since cutting the film piece into the desired shape can affect the dose response properties of the film.
<table>
<thead>
<tr>
<th>Scan number</th>
<th>Mean pixel value</th>
<th>Standard deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>216</td>
<td>2</td>
</tr>
<tr>
<td>2</td>
<td>215</td>
<td>1</td>
</tr>
<tr>
<td>3</td>
<td>215</td>
<td>3</td>
</tr>
<tr>
<td>4</td>
<td>216</td>
<td>2</td>
</tr>
<tr>
<td>5</td>
<td>214</td>
<td>2</td>
</tr>
<tr>
<td>6</td>
<td>215</td>
<td>2</td>
</tr>
<tr>
<td>7</td>
<td>217</td>
<td>4</td>
</tr>
<tr>
<td>8</td>
<td>215</td>
<td>2</td>
</tr>
</tbody>
</table>

*Table 2.3: The mean pixel value for an unexposed 8 x 80 mm GafChromic film piece placed at different positions on the scanner bed.*

From the results in table 2.3 it can be concluded that the scanner position dependence is low, with the largest variation found in scan number 7 with a standard deviation of 1.8% and a total standard deviation for all the measured grey values of 0.4%. A Kolmogorov-Smirnov statistical test was done in order to examine the probability that the different distributions actually come from the same total distribution that is valid for the entire scanner. The test gave a probability of approximately 80% that the different distributions actually originate from the same total distribution for the entire scanner bed.

To examine the whole-area uniformity of the scanner two different scans were taken of the entire scanner bed, one for the entire scanner and one for the entire scanner bed but excluding the areas within 2 cm from the scanner bed edges. The pixel value histograms of these scans were then examined. This test is important since a non-uniform light source will affect the readings and cause erroneous results.
The whole-area uniformity test showed that the total scanner bed uniformity is very good with a standard deviation of only 0.1% when the outer 2 cm of the scanner bed are excluded. Since the standard deviation is 4.7% when the edges are included this means that the scanner is much more uniform in the central areas of the scanner bed with the standard deviation being almost 50 times lower. Therefore all the film pieces should be scanned in the central areas of the scanner to avoid non-uniformity errors.

2.3.3 Dependence on acquisition window geometry

The area that is scanned for each measurement can be defined manually using the control software. A small area will of course result in less disk space and shorter scan time required for each measurement, since the part of the scanner bed that is not used for evaluating the film piece is ignored. A check was done to see whether the size of the chosen scanning area affected the outcome of the measured pixel value or not. To do this, two unexposed pieces of film were placed on the scanner bed in different positions and scanned with different sizes on the scanning area.
A total of five scans were made, one with both pieces close to one another (pos1), one with the film pieces separated in the left-right position and with a small scanning area (pos2, Fig 2.10), one with the same separation between the film pieces in the left-right position but with a larger scanning area (pos3, Fig 2.10), one with a separation in the up-down position and a small scanning area (pos4 Fig 2.10) and finally one with the same separation in the up-down position but with a larger scanning area (pos5 Fig 2.10).

<table>
<thead>
<tr>
<th>Position</th>
<th>Grey value piece1</th>
<th>Grey value piece2</th>
<th>Difference [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>205.5</td>
<td>204.8</td>
<td>0.3</td>
</tr>
<tr>
<td>2</td>
<td>216.9</td>
<td>212.1</td>
<td>2.3</td>
</tr>
<tr>
<td>3</td>
<td>207.5</td>
<td>206.1</td>
<td>0.7</td>
</tr>
<tr>
<td>4</td>
<td>205.4</td>
<td>207.5</td>
<td>1.0</td>
</tr>
<tr>
<td>5</td>
<td>209.0</td>
<td>207.7</td>
<td>0.6</td>
</tr>
</tbody>
</table>

Table 2.4: The measured grey values for two unexposed GafChromic film pieces from the same sheet. The different positions are shown in figure 2.10.

The results showed that the measured pixel values (that later are converted to absorbed dose) can differ up to 2.3 % for the two film pieces if they are placed in the left-right position and the scanning area is chosen to exactly match the required area (pos2, Fig 2.10). If the film pieces are placed close to each other and the scanning area is large (pos1, Fig 2.10) the difference in pixel value is only 0.3%. If the pieces are separated and scanned in the up-down position of the scanner and the scanning area is chosen to be large, the difference in the pixel values and thus in the absorbed dose is approximately 0.6 % (pos5, Fig 2.10). A good result is also achieved if the pieces are
scanned in the left right direction with a large scanning area with a difference of 0.7% (pos.3 Fig 2.10).

Therefore, where possible, the film pieces should be placed close to each other and with the scanning area chosen to be significantly larger than the required scanning area since this will yield the lowest difference in the mean pixel value and thus in the absorbed dose between identical pieces. When many pieces are to be scanned and the pieces can not all be placed within a small area, the film pieces should be placed in the up-down position of the scanner bed and with the scanning area significantly larger than the required scanning area (pos5 figure 2.10). In this study, since many pieces are always scanned at once, the film pieces are always scanned in the up-down position of the scanner bed and with a large scanning area.

2.4 Test for repeatability of film values after multiple scans

To ensure that it is enough to scan each piece once for reliable results, and since some of the GafChromic film pieces are scanned multiple times, it is important to know if the obtained pixel values are reproducible from scan to scan. To check this a number of film pieces that had been exposed to different doses were scanned multiple times and the mean pixel value from each scan was compared.

Table 2.5 shows the mean pixel values from three different scans for three different GafChromic film pieces. The standard deviation in the mean values for the three different scans is 0.2 % for all pieces. This variation is small enough not to affect the measured results and therefore one scan of every piece of film is sufficient.

<table>
<thead>
<tr>
<th>Dose</th>
<th>Scan 1</th>
<th>Scan 2</th>
<th>Scan 3</th>
<th>Mean</th>
<th>Std [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 Gy</td>
<td>210.1</td>
<td>209.3</td>
<td>210.1</td>
<td>209.8</td>
<td>0.2</td>
</tr>
<tr>
<td>6 Gy</td>
<td>138.1</td>
<td>137.6</td>
<td>137.6</td>
<td>137.8</td>
<td>0.2</td>
</tr>
<tr>
<td>12 Gy</td>
<td>98.3</td>
<td>97.9</td>
<td>97.9</td>
<td>98.1</td>
<td>0.2</td>
</tr>
</tbody>
</table>

*Table 2.5 Table showing how the measured pixel values changes if the film pieces are scanned multiple times.*
2.5 Energy dependence and derivation of a dose-NOD calibration curve

Ten pieces of GafChromic MD55 film and fourteen pieces of HS film (approximately 1x1 cm$^2$) were cut out from two different sheets and all pieces except two were irradiated uniformly by $^{192}$Ir in a special built perspex phantom (figure 2.11). The two unexposed pieces were kept to use for calculations of the NOD. The other pieces were given the following doses:

Two pieces at each dose level from 1 Gy to 4 Gy for the MD55 film pieces.
Two pieces at each dose level from 1 Gy to 6 Gy for the HS film pieces.

Thereafter one film piece from each dose level was given an additional dose with 6MV photons produced in a linear accelerator. The reason for exposing the film to both 6MV photons and $^{192}$Ir energies is to investigate the energy response of the film. The extra dose given was 4 Gy for the HS film pieces and 5 Gy for the MD55 film pieces. Thus the absorbed dose for the GafChromic MD55 film pieces ranges between 0 - 9 Gy and between 0-10 Gy for the HS film pieces, with the low doses given by $^{192}$Ir and the high doses given by both $^{192}$Ir and 6 MV photons (Table 2.6). The reason why the dose interval is chosen to be between 0-10 Gy is because the MD55 and HS films have a nearly linear relation between NOD and absorbed dose in this interval when used together with the Hewlett Packard ScanJet II cx/T scanner.

<table>
<thead>
<tr>
<th>Radiation type</th>
<th>GafChromic MD55</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dose [Gy]</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>*</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Radiation type</th>
<th>GafChromic HS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dose [Gy]</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>*</td>
</tr>
</tbody>
</table>

Table 2.6: The dose given to the GafChromic film pieces in order to examine the film’s energy dependence. * The film is unexposed. ** The dose is given by 6 MV photons. For the MD55 pieces the dose between 1-4 Gy is given by $^{192}$Ir, 5 Gy is given with only 6 MV photons and 6-9 Gy is given by both $^{192}$Ir and 6 MV photons. For the HS film the dose 1-6 Gy is given by $^{192}$Ir, 4 Gy only by 6MV photons and 4-10 Gy by both $^{192}$Ir and 6 MV photons.
Figure 2.11: The special built phantom used for irradiation of the Gafchromic film pieces. The $^{192}$Ir source dwelling positions are calculated by a treatment-planning program in order to achieve an uniform dose distribution to the GafChromic film pieces. Two film pieces are irradiated at the same time.

Since the film pieces were taped on the outside of the perspex tube through which the catheter with the source runs, the distance to the film pieces is known. Therefore a treatment plan can be made in order to get an uniform dose distribution at that distance. Figure 2.12 shows the dose distribution calculated with the treatment planning system. The squares represent the source dwelling positions and the dotted line represents the desired dose that covers the film uniformly. The film is represented as dots in the figure.
The film pieces that were exposed to an additional dose with 6 MV photons were placed in a solid water phantom and given half of the intended dose, thereafter the pieces were shuffled around within the Field Of View (FOV) and given the rest of the dose. By shuffling around the pieces within the FOV the effect of any fluence inhomogeneity that might occur in the photon beam is minimised.

The solid water phantom used in this experiment is shown in figure 2.13. The film pieces are placed between a 1.5 cm thick slab of solid-water and a 5.15 cm slab of solid water at an SSD of 100 cm. The top solid-water slab is needed in order to get full build up for the photons and the bottom slab is used to get the appropriate backscatter.
Figure 2.13: The set-up used for irradiation of the GafChromic film pieces that are used to obtain the calibration curves for the different film sheets. The top solid-water slab produces full build up conditions and the bottom slab is required in order to get full backscatter to the film pieces.

The next step is to find a relation between the NOD and the absorbed dose. To do this, the measured NOD is plotted as a function of the absorbed dose to the film pieces and a straight line is fitted to the plotted points. The straight line is forced to go through the point (0,0) since zero dose gives zero NOD according to the definition of the NOD. The equation of the straight line will give the desired relation between NOD and absorbed dose for that specific sheet. This relation between NOD and absorbed dose is called the calibration curve for the film and has to be known for every film sheet used since the sensitivity and the dose response from sheet to sheet may vary.

In the figures below the NOD as a function of absorbed dose to the film pieces is shown for the MD-55 and the HS film pieces. The linear fit suggests an approximately linear relation, with a correlation coefficient of 0.998 for the MD55 model and 0.999 for the HS model, in the interval 0 - 10 Gy, even though the doses were given with different energies and different modalities. This means that 6 MV photons produced in a linear accelerator can be used for irradiation of the calibration curve film pieces even though the measured depth doses are obtained with $^{192}$Ir energies without any significant difference in the results. Since there is no significant difference in energy response for 6 MV photons and $^{192}$Ir energies, a couple of the calibration curve film pieces used in this thesis are irradiated using only 6 MV photons. Doing so will save time and work since the experimental set-up for exposure with $^{192}$Ir is more complicated and the calibration
procedure is more time consuming. In this particular case the calibration curves suggest the following relations between NOD and absorbed dose.

Dose = NOD/0.0274 for the GafChromic MD55 film.

Dose = NOD/0.0523 for the GafChromic HS film.

Figure 2.14: Ten GafChromic MD-55 film pieces were exposed to doses in the interval between 0-9 Gy. The film pieces were exposed with both $^{192}$Ir and 6MV photons. A linear fit to the data points gives a correlation coefficient of 0.998.

Figure 2.15: Fourteen GafChromic HS film pieces were exposed to doses in the interval between 0-10 Gy. The film pieces were exposed with both $^{192}$Ir and 6MV photons. A linear fit to the data points gives a correlation coefficient of 0.999.
2.6 Post exposure density change

The rate of the post exposure change in optical density is important to know since the different irradiations and measurements are done at different times. With the knowledge of the rate of change of optical density, the measured results can be corrected if the time of irradiation for the film pieces differs from the time of irradiation for the calibration curve. The pixel values of a number of MD55 pieces that had been exposed to different doses were measured at different times after irradiation and plotted as a function of time in order to investigate the post exposure effect on the film.

![Figure 2.16: The change in grey value with time for GafChromic MD55 film pieces exposed to different doses.](image)

The results show that the change in optical density after irradiation is linear and constant within ± 14% for doses between 3 - 18 Gy and in the time interval of 48 - 550 hours after exposure. The mean change rate for doses in the range of 3 - 18 Gy is 0.012 pixel values per hour with a standard deviation of 0.002. This corresponds to a change in dose of approximately 3 cGy per 24 hours. Even though the standard deviation is approximately 14%, the mean value of the change rate can be used for all doses since the time between the depth dose irradiations and the irradiations of the calibration strip...
is only a couple of hours. Therefore the difference in optical density change is very small and the errors due to a small difference in the rate of change for different doses are insignificant.

The pixel value for the MD55 film model at a certain time after irradiation can thus be calculated with the following formula:

\[
\text{Pixel Value (t)} = \text{Pixel Value (48)} - 0.0123 \times (t - 48)
\]

where \( t \) is the time after exposure in hours and \( \text{Pixel Value (48)} \) is the measured pixel value 48 hours after exposure.

### 2.7 Film scanning protocol

As mentioned in this chapter there are a number of steps to follow in order to calculate the absorbed dose to a film piece. The first step is to record a transmission image for the irradiated film piece. This is done by using a normal flat-bed scanner in transmission mode and scanning the film piece. The scanned image is saved as a million colours TIF-file.

The obtained image is then split into the red, the blue and the green colour channels using a photo editing program (Photostyler 2.0) and thereafter the red colour channel image is exported to an image analysing program (Image J). The reason for splitting the image into different colour channels and only using the red one for analysis is because the major absorption peak is located in the red part of the spectrum for the GafChromatic film. This means that using light with the wavelength in the red part of the spectrum to measure the transmission through the film will result in a higher sensitivity.

The image analysing program (Image J) is then used to measure the pixel values in different parts of the film. The pixel values are proportional to the intensity of the light transmitted through the film, a high pixel value corresponds to a high intensity of the transmitted light thus a low absorbed dose. Since the pixel value is proportional to the intensity of the transmitted light it can be used to calculate the NOD using equation 2.2 and 2.3 with the pixel value inserted instead of the intensity \( I \).
In order to minimise scanning errors, each time a film piece was scanned for analysis it was scanned together with the irradiated pieces used to obtain the calibration curve for the actual sheet. Thus each scan has its own calibration curve even if the film pieces are from the same film sheet.

Another important thing that has to be considered for each case is the post exposure correction. If the film pieces used for the calibration curve were irradiated at a different time than the film pieces used to measure dose under the Leipzig applicators, either the calibration curve pieces or the pieces used for dose profile measurements under the applicators have to be corrected for the post exposure density growth of the films.

![Image](image.png)

*Figure 2.17: The procedure used to obtain the absorbed dose from the scanned images. 1. The exposed film piece is scanned in transmission mode 2. The red colour channel information is extracted and used to measure the NOD 3. The calibration curve for the scan is plotted and the relation between NOD and absorbed dose is given by the slope of the fitted line 4. The absorbed dose at every point can be calculated using the relation obtained from the calibration curve.*
3 Theory and dose modelling

3.1 The TG43 dose calculation formalism

In 1988 the American Association of Physicists in Medicine (AAPM) formed the task group number 43 (TG43) to review the publications of interstitial brachytherapy sources and recommend a dosimetry protocol that includes a formalism for dose calculations. This dosimetry protocol is based on the unit air kerma strength instead of the exposure rate constants that were used earlier and could differ quite significantly for the same radioisotope. The dosimetry protocol was first published in 1995 (Nath et al. 1995), then revised and updated in 2004 (Rivard et al. 2004). The main reasons for the update were that the utilisation and the number of low-energy interstitial brachytherapy source models available had increased in this time and that the National Institute of Standards and Technology (NIST) had introduced a new primary standard of the air kerma strength. Furthermore, the dosimetry methodologies and dosimetric characterisations of particular source models had improved. Therefore the AAPM considered that an update was necessary.

The TG43 protocol suggests the following equation for obtaining the dose rate at a point P(r,θ) at the distance r and the polar angle θ from a cylindrically symmetric line source, assuming that the source is centred at the origin:

\[
D(r, \theta) = S_K \cdot \Lambda \cdot \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} \cdot g_L(r) \cdot F(r, \theta)
\]  

(3.1)

\(r\) is the distance from the centre of the active source to the dose point P.
\(r_0\) is the distance to the reference point P(\(r_0, \theta_0\)) which is 1 cm in the TG43 protocol.
\(\theta\) is the polar angle for the point P relative to the source longitudinal axis.
\(\theta_0\) is the polar angle to the reference point, which is specified to be 90°.
\(S_K\) is the air kerma strength of the source \([cGy \cdot cm^{-2} \cdot h^{-1}] = [U]\)
\(\Lambda\) is the dose rate constant in water. \([cGy \cdot h^{-1} \cdot U^{-1}]\)
\(G(r,\theta)\) is the geometry function at the distance \(r\) and the polar angle \(\theta\).
\(g(r)\) is the radial dose function at the distance \(r\) from the origin.
\(F(r,\theta)\) is the anisotropy function at the distance \(r\) and the polar angle \(\theta\).
Figure 3.1: Notations used by the TG43 dose calculation formalism for brachytherapy sources. The source is assumed to be surrounded by water. The angle $\beta$ is the angle subtended by the active part of the source. $\beta = \theta_2 - \theta_1$

The suggested equation (3.1) assumes that the dose distribution is cylindrically symmetric with respect to the longitudinal axis of the source but can be generalised to use for non-symmetric sources as well.

The air kerma strength, $S_K$, is numerically identical to the reference air kerma rate recommended by ICRU 38 and ICRU 60. The unit of $S_K$ is $\mu$Gy m$^2$ h$^{-1}$ and is sometimes denoted by U, which stands for unit air kerma. $S_K$ is defined as the air kerma rate $\dot{K}_\delta(d)$ in vacuum at the distance $d$ from the source centre multiplied by the square of the distance. $S_K = \dot{K}_\delta(d) \cdot d^2$.

The distance $d$ above can be chosen to be any distance as long as it is large in comparison to the source dimensions and the detector so that the geometry of the source can be accurately represented by an ideal point in space.

Air kerma rate in vacuum means that it is corrected for photon attenuation and scattering if the measurements were done in air. The cut off energy $\delta$ should be chosen so that the low energy photons that contribute to the air kerma rate but not significantly to doses at depths larger than 1mm are excluded. The cut off energy is often chosen to be 5 keV for low-energy photon-emitting brachytherapy sources.
The dose rate constant in water $\Lambda$ is the ratio of the dose rate at the reference point $P(r_0,0_0)$ and the air kerma strength in water.

$$\Lambda = \frac{D(r_0,\theta_0)}{S_K} \quad [\text{cGy h}^{-1} \text{U}^{-1}] \quad (3.2)$$

The dose rate constant is dependent on the radionuclide type and the design of the source model used. This constant includes the effects of source geometry, spatial distribution within the source of the source activity, encapsulation, self-filtration within the source and scattering in the water surrounding the source.

The geometry function suppresses the steepness in tabulated dose data and thereby improves interpolation accuracy. Since the geometry function is only used for interpolating between different discrete values one can use a simple approximation of the geometry and still have sufficient accuracy in the calculated values. The TG43 protocol recommends using a line source approximation for the two-dimensional case i.e. for the cylindrical symmetric line source. According to the updated version of the AAPM TG43 report the geometry factor is given by:

$$G = \begin{cases} \frac{\beta}{L r \sin \theta} & \text{if } \theta \neq 0^\circ \\ (r^2 - L^2 / 4)^{-1} & \text{if } \theta = 0^\circ \end{cases} \quad (3.3)$$

where $\beta$ is the angle in radians (see Fig 3.1)

The radial dose function $g_L(r)$ corrects for the dose fall off due to attenuation and scattering of the photons in the transverse plane. The radial dose function is given by

$$g_L(r) = \frac{D(r,\theta_0)}{D(r_0,\theta_0)} \cdot \frac{G_L(r_0,\theta_0)}{G_L(r,\theta_0)} \quad (3.4)$$
The anisotropy function for the cylindrical line source is defined as

\[
F(r,\theta) = \frac{D(r,\theta)}{D(r,\theta_0)} \frac{G_z(r,\theta_0)}{G_z(r,\theta)}
\]  
(3.5)

This function describes how the dose varies with the change in the polar angle relative to the transverse plane. It includes the effects of self-filtration, oblique filtration of the primary photons through the encapsulating material and scattering of the photons in the medium. The anisotropy is defined as unity on the transverse plane and decreases as \( r \) decreases, as \( \theta \) approaches 0° or 180°, as the encapsulation thickness increases or as the photon energy decreases.

The TG43 dose calculation model assumes that the radioactive material in the source is only surrounded by water and a melded steel capsule inside which the radioactive material is contained. This assumption is important to keep in mind since this is not the case for the Leipzig applicators. For the Leipzig applicators one has a steel tube that is melded onto the applicator inside of which the source travels, an air gap between the source and the treatment surface, the plastic protective cap and the tungsten/steel collimators positioned around the source. Each of these factors will affect the dose distribution in different ways and therefore the dose distributions obtained with the TG43 model will not completely agree with the real dose distributions beneath the Leipzig applicators.

In order to investigate the combined effect of these factors on the dose distribution, the obtained dose profiles with the TG43 model are compared with Monte Carlo simulated profiles that explicitly takes these complex interactions into account. The Monte Carlo profiles that are used for comparison have been derived by Perez et al. (2005).

The TG43 profiles for a single source capsule completely surrounded by water was obtained by interpolating the table given by Daskalov et al. (1998). In this study, MC photon transport simulation techniques have been used to characterise the different factors that are included in the TG43 dose calculation formalism. The results for each contributing factor are presented in their study and the dose-rate distribution around the source is tabulated in a 2D along-and-away look-up table. Each point is thus described with an along (\( y \)) and an away (\( x_{\text{real}} \)) coordinate. To use this data, each point beneath the
$^{192}$Ir source has to be described in the $x_{\text{real}}$ and $y$ coordinate system where the $y$ axis is parallel and the $x_{\text{real}}$ axis is perpendicular to the source's longitudinal axis, both with the origin at the source centre (Fig. 3.2). The $x_{\text{real}}$ value is the distance away from the source along the transverse plane and is given by

$$x_{\text{real}} = \sqrt{x^2 + z^2} \quad (3.6)$$

where $x$ and $z$ are the values along the $x$ and $z$ axes as shown in figure 3.2.

Since the $^{192}$Ir source is surrounded by the tungsten/steel alloy collimators, scatter from such high-Z materials have to be taken into account. For simplicity, the collimator effects are assumed to be a step function, so that the collimator factor is unity inside the treatment field, but zero outside. The treatment field is derived by extending an imaginary line from the collimator surface into the water phantom (the dashed lines in Fig 3.2).

For a profile at a certain depth for the vertical applicator the $y$ coordinate is given by the depth beneath the skin surface plus the distance from the skin surface to the centre of the source. The $x_{\text{real}}$ coordinates ranges from zero at the central axis to $d$, where $d$ is the distance from the central axis to the collimator wall at that specific depth. The distance $d$ to the collimator wall at a certain depth is given by

$$d = (\text{depth} + 1.6) \cdot \tan \alpha \quad (3.7)$$

where $\alpha$ is the collimator angle as shown in figure 3.2. Since the profile is symmetrical around the central axis the other half of the profile, from the central axis to $-d$, is the same as the profile from the central axis to $d$. 
Figure 3.2: The coordinate system used for calculation of the TG43 profiles.

The principle for obtaining the profiles for the horizontal applicators is similar to the one used for the vertical applicators. However, the dose distribution is no longer symmetrical around the central axis since the wire that drives the source into position perturbs the dose distribution on the proximal or inferior end of the source. Therefore two different profiles are calculated for each applicator, one parallel with the source and one perpendicular to the source. Figure 3.2 shows the principle used to get the x and y coordinates for the horizontal case.

The dose rate table presented by Daskalov et al. (1998), only contains values for certain points. The values in between are calculated by using linear interpolation between the given points in the table. Figure 3.3 shows TG43 calculated profiles for the V20 and H20 applicators at skin surface. As can be seen, multiplying the dose rate by the collimator function causes the dose outside the collimator to fall to zero. This can not be the case for the real dose distribution, where scatter in the phantom and from the collimator walls must yield a steep dose penumbra in this region.
Figure 3.3: The dose rate profiles at skin surface for the V20 (left-hand figure) and H20 (right-hand figure) applicators at skin surface. The profiles have been calculated using the TG43 dose calculation model.

To illustrate the difference between the TG43 model and the Monte Carlo simulations and investigate the effects of the Leipzig applicator on the dose distribution, the two profiles are compared for the H20 and V20 applicators at three different depths ranging from 0.1 cm to 0.5 cm beneath the skin surface.
Figure 3.4: The TG43 dose profiles are compared with the Monte Carlo profiles in order to investigate the effects of the cone collimators, the air gap, the protection cap and the steel tube on the dose distribution. The profiles are shown for the V20 and H20 applicators. The mean difference shows how much the TG43 profile differs from the Monte Carlo profile. The errors in the Monte Carlo profiles are less than 0.5 % and for the TG43 model the errors range from 0.5 % close to the source to 1.2 % far from the source.

To understand and explain the differences between the two models, the different interactions that occur for the two different set-ups have to be investigated.

The characteristic electrons that are emitted from the $^{192}$Ir source have very low energies (70 - 210 keV) and are therefore easily absorbed in the encapsulation layer. For the TG43 model they are assumed to be absorbed either in the source encapsulation or in the surrounding water, and in the Monte Carlo model they are absorbed either in the steel tube surrounding the source or the plastic protective cap. Therefore, the characteristic electrons emitted by the source will not contribute to the profiles at the depths that we are calculating. The characteristic photons will however reach these depths and interact on the way producing secondary photons and electrons. The most
obvious difference that can be seen in Fig 3.4 at the surface is the difference in dose rate around the central axis.

Here the TG43 model predicts a higher dose rate than the Monte Carlo model. The assumption made by the TG43 model is that the source is surrounded by water and therefore full scatter conditions exists everywhere. This assumption can not be valid near the surface because there is a lack of forward scatter of secondary particles from the air layer above. Only the Monte Carlo model accurately describes the asymmetric scatter conditions at the surface. However, as depth increases, full scatter conditions apply and this appears consistent with the above two models converging at 5 mm depth. This can be seen as the difference in dose rate close to the central axis between the two models decreases with depth.

Another difference between the two models is the dose rate close to the edges of the collimators. The high-Z material in the collimators is effective at scattering photons, these scattered photons are able to penetrate as far as 5 mm in water. However, the TG43 model is unable to model this interaction, hence it reports a lower dose rate than the Monte Carlo model. The contribution of the scattered photons should be larger close to the collimator walls and become less dominant near central axis.

Each photon-matter interaction causes the photon to lose some of its energy or disappear in a photoelectric effect. A photon with lower energy has a higher probability of interaction than a high-energy photon and will therefore be absorbed faster. This means that photon beams with lower mean energies, like the scattered photons, will not penetrate as deep into a medium as photon beams with higher energies, like the primary photons. The differences between TG43 and Monte Carlo due to scattered radiation from the collimator walls appear to diminish as the depth increases, since there are fewer scattered photons remaining at larger depths. The absorption of the scattered photons from the collimator walls and the increase of scatter from water as depth increases are the reasons to the decrease in the mean difference between the two models as the depth increases. For the H20 applicator the mean difference decreases from 2.17 % at 0.1 cm depth to 0.93 % at 0.5 cm depth and for the V20 applicator the mean difference decreases from 2.89 % at 0.1 cm depth to 1.75 % at 0.5 cm depth. The above effects are also found to be consistent for the other applicators as well.
3.2 The equivalent water distance TG43 model

As described in the previous section there are a number of factors that affect the dose distribution beneath the Leipzig applicators that are not taken into account by the TG43 model. This could be seen as the profiles obtained with the TG43 model differed from those obtained with the Monte Carlo model. In order to try to improve the agreement of the TG43 model an attempt was made to modify it. Since the TG43 calculation formalism assumes that the cylindrical $^{192}$Ir source is only surrounded by water a new model that takes the steel tube that is melted onto the applicator and the air gap into account was made. The model is based on the same principles as the original TG43 dose calculation formalism but it uses the equivalent water distance to a certain point rather than the real distance for evaluation of the anisotropy function $F(r, \theta)$ and the radial dose function $g(r)$. For evaluation of the geometry factor the real distance to the point is still used since the material surrounding the source does not affect the geometry factor. The distance to a certain point $P$ can be divided into three parts, a distance through the steel tube, $d_{\text{steel}}$, a distance through air, $d_{\text{air}}$, and a distance through water, $d_{\text{water}}$. The equivalent water distance, $d_{\text{eq}}$, is then calculated using the following relation:

$$d_{\text{eq}} = \frac{d_{\text{water}} \cdot \mu_{\text{water}} \cdot \left( \frac{\mu}{\rho} \right)_{\text{water}} + d_{\text{air}} \cdot \mu_{\text{air}} \cdot \left( \frac{\mu}{\rho} \right)_{\text{air}} + d_{\text{steel}} \cdot \mu_{\text{steel}} \cdot \left( \frac{\mu}{\rho} \right)_{\text{steel}}}{d_{\text{water}} \cdot \mu_{\text{water}} \cdot \left( \frac{\mu}{\rho} \right)_{\text{water}}} \tag{3.8}$$

where $d_{\text{total}}$ is the real distance between the source centre and the point $P$.

This way of calculating the equivalent water depth takes into account that a considerable fraction of the particles interact via the photoelectric effect, especially in the steel tube. The mean energy of the $^{192}$Ir was used, 400 keV, to obtain the mass attenuation coefficients for the different materials. In fact, the mean energy of $^{192}$Ir is 370 keV but since most of the low energy photons are absorbed by the steel tube surrounding the $^{192}$Ir source you get some beam hardening and therefore a mean energy of 400 keV is more suitable in the case of the Leipzig applicator. By using this equivalent water model of the original TG43 model, the steel tube and the air gap is taken into account. A schematic figure of the steel encapsulated source is shown in figure 3.5.
Figure 3.5: Schematic figure of the vertical $^{192}$Ir source with the steel tube surrounding the source, the air gap and the water. The total distance to the point P is divided into three different distances, $d_{\text{steel}}$, $d_{\text{air}}$, and $d_{\text{water}}$. The distances are used to calculate the equivalent water distance to the point according to equation 3.8.

The thickness of the front steel wall, $d_{\text{wall}}$, was not known and could not be measured, therefore this distance is chosen so that the fit between the two models is as good as possible. The best fit was found using a thickness of $d_{\text{wall}} = 2.4$ mm.

The same principle that was used for the vertical applicators was also used for the horizontal applicators. The schematic figure used for calculations of the equivalent water depth is shown in figure 3.6.
Figure 3.6: Schematic figure for the horizontal incoming $^{192}\text{Ir}$ source. The total distance to the point P is divided into $d_{\text{steel}}$, $d_{\text{air}}$, and $d_{\text{water}}$ which are then used to calculate the equivalent water distance to that point. The equivalent water distance is then used in the TG43-EQ model to obtain the anisotropy and the radial dose function.
Figure 3.7: The TG43-EQ calculated dose profiles compared with the Monte Carlo calculated profiles for the V20 and H20 applicators. The two models are compared at three different depths and for each profile the mean difference and the max difference from the Monte Carlo model is showed. The errors in the Monte Carlo profiles are less than 0.5 % and for the TG43 model the errors range from 0.5 % close to the source to 1.2 % far from the source.

Figure 3.7 shows the TG43-EQ profiles compared with the Monte Carlo profiles for the V20 and H20 applicators at the same depths as before. The same discussion as before is valid even in these cases with the dose rate being higher close to the central axis for the TG43-EQ model than for the Monte Carlo model due to the full scatter conditions assumed by the TG43 model and the dose rate being lower for the TG43-EQ model close to the collimator edges due to collimator scatter. The differences are however smaller than for the TG43 model since the steel tube and the air gap has been taken into account. Table 3.1 shows how the mean difference changes for a couple of profiles when using the TG43-EQ model instead of the ordinary TG43 model.
Taking the steel tube and the air gap into account thus gives a slightly better agreement with the Monte Carlo profiles. To obtain an even better agreement the scatter from the collimator walls has still to be taken into account.

Furthermore, the dose outside the collimator walls has not been modelled with the TG43, nor the TG43-EQ models but rather ignored. This will however not have a significant effect in this study since we are only interested in the dose distribution within the treatment area. It could be done in order to make the TG43 model even more accurate for the Leipzig applicators.

A general conclusion that can be made from the results in this chapter is that using the TG43-EQ model generally gives a better agreement with the Monte Carlo model than using the ordinary TG43 model. The TG43-EQ model that was compared in this chapter only took into account the steel tube and the air gap between the source and the treatment surface. By modifying it even more to take the scatter from the collimator walls into account the agreement with the Monte Carlo model would probably be even better. The errors in the Monte Carlo data are less than 0.5 % and for the TG43 model the errors range from 0.5 % close to the source to 1.2 % far from the source (Daskalov et al. 1998, Perez et al. 2005) meaning that most of the profiles are within the error margins.

The good agreement between the TG43-EQ and the MC models for the different applicators suggests that the TG43 models can be used for quick checks of the depth dose distribution and verification of the treatment plans for the Leipzig applicators. Doing so, it is important to keep in mind that using the TG43-EQ model results in mean

Table 3.1: This table shows the mean differences between the Monte Carlo profiles and the profiles that are obtained with the TG43 and TG43-EQ models for the Leipzig applicators at different depths. The agreement with the Monte Carlo model is generally better for the TG43-EQ model.
errors smaller than 2.3 % for the vertical applicators and 2.0 % for the horizontal applicators compared to using Monte Carlo simulations.
4 Dose measurements

4.1 Dose distribution of Leipzig cones

In this study the dose distribution at skin surface, at 3.38 mm and at 5.19 mm equivalent water depth has been measured for all six applicators. The water equivalent depths at which the dose distributions were measured are obtained by using the following formula:

\[ d_{eq} = \frac{d \cdot \rho}{\rho_{water}} \]

where \( d \) is the actual thickness of the solid-water or PMMA slab used to simulate a certain depth and \( \rho \) is the density, which for solid-water is 1.02 g/cm\(^3\) and 1.19 g/cm\(^3\) for the PMMA slab. The density for water, \( \rho_{water} \), is 1.00 g/cm\(^3\).

For each measurement an appropriate film piece was cut into shape and placed at the appropriate depth. Thereafter the applicator was put into position and held down with a metal clamp in order to ensure that the applicator surface was parallel with the treatment surface and that the applicator could not move during irradiation (Fig 4.1). The first measurements that were performed showed a non-symmetrical dose distribution for the vertical applicators (Fig 4.2). This asymmetry was observed reproducibly for all the measurements done for the vertical applicators even when the experimental set-up was modified in the attempt to measure a more symmetric profile. This was thought to be due to a small tilt in the source position meaning that the \(^{192}\)Ir source was not perpendicular to the treatment surface. A mean measured profile was therefore used for evaluating the measurements for the vertical applicators.
Figure 4.1: Experimental set-up for measuring the depth dose profiles. The perspex or solid-water slab simulates a certain water equivalent depth and the bottom Solid Water slab is used to get the appropriate backscatter conditions. The left-hand figure shows the set-up for measurements at the skin surface and the right-hand figure shows the set-up for measurements at a certain depth.

Figure 4.2: The measured dose profiles at skin surface for the three vertical applicators. a) The V20 applicator at skin surface, b) the V10 applicator at skin surface and c) the V30 applicator at skin surface.

The mean profile is calculated by averaging the data points from a circle with radius \( r \) from the central axis, into one data point and plotting these mean value data points into
a dose profile (Fig 4.3). The mirror image of the obtained mean profile is then used for the other half of the profile and thereafter the whole profile can be compared with the TG43 calculated data. This procedure is used for all the vertical applicators. The use of the mean profile instead of the real measured profile will of course give rise to errors in the measured data, but since it is very difficult to assure that the incoming $^{192}$Ir source is vertical relative to the treatment surface this is the best method to use in order to get a symmetrical dose distribution around the central axis.

4.1.1 Vertical applicators

The mean measured and the TG43-EQ profiles are shown for the V20 and V30 applicators at skin surface and at 5.19 mm depth in figure 4.4. The differences between the TG43-EQ model and the measured profiles are due to the same reasons as the differences between the TG43-EQ and the Monte Carlo profiles described previously in chapter 3. Once again the TG43-EQ model predicts a higher dose rate close to the central axis due to the assumption that full scatter conditions apply even at skin surface. The influence of the scatter from the collimator walls is visible as well, as the decrease in dose rate is faster for the TG43-EQ model as one move away from the central axis. As depth increases the difference between the TG43-EQ and the measured profiles decreases as full scatter conditions begin to apply for the measured profiles and since

Figure 4.3: The mean value of the data points from a circle with radius $r$ from the central axis is plotted as one point in the mean value dose profile. Thereafter the mirror image of the obtained profile is used for the other half of the profile giving a symmetrical profile that can be compared with the calculated TG43 profile.
much of the scattered radiation from the collimator walls has been absorbed. This is again consistent with the discussion made in chapter 3.

The errors in the TG43-EQ data range from 0.5% close to the source to 1.2% far from the source (Daskalov et al. 1998, Perez et al. 2005). The errors in the measured data are 1.2% due to errors from the scanning procedure (0.8% due to the scanner position dependence and 0.4% due to the dependence of the acquisition window size) and up to 4% due to the uncertainty in the GafChromic MD55 film's response. The errors in the measured absolute dose thus add to approximately 5% meaning that the measured profiles agree within the error margins with the TG43-EQ profiles. An additional error source is the tilt in the source position. Using a mean profile instead for the actual profile changes the shape of the dose distribution and therefore affects the agreement with the calculated profile since the tilt angle is not taken into account by the TG43-EQ model.

Figure 4.4: The mean measured profile is compared with the TG43-EQ calculated profiles for the V20 and V30 applicators at skin surface and 5.19 mm water equivalent depth. The errors in the TG43-EQ are approximately 1.2% and the uncertainty in the measured results is up to 5%.
4.1.2 Horizontal applicators

A comparison between the TG43-EQ profiles and the Monte Carlo profiles was also done for the horizontal applicators at the same depths as for the vertical applicators. Since the profiles are not symmetrical around the central axis of the source, two profiles were compared at each depth. One profile parallel with the source's longitudinal axis and one profile perpendicular to the source's longitudinal axis. However, only the profiles perpendicular to the source are shown since the parallel profiles are very similar and do not present any additional information.

Figure 4.5: The mean measured profile (perpendicular to the source) compared with the TG43-EQ profile for the H20 applicator at a) skin surface, b) 3.38 and c) 5.19 mm depth. The uncertainty in the measured profiles is approximately 5%.
Once again the mean difference between the two profiles decreases as the depth increases due to the fact that the scattered radiation from the collimator walls is absorbed and since full scatter conditions start to apply for the measured profiles at depth. However, there is a noticeable difference from the previous comparisons. At skin surface and at 3.38 mm depth, the TG43-EQ model predicts a continuously lower dose rate throughout the entire profile. This continuously lower dose rate predicted by the TG43-EQ model has been seen for a number of profiles when compared with actual measurements. It is hard to say what this depends on without further measurements but one hypothesis is that it could depend on a small displacement in the source position, meaning that the source is actually closer to the treatment surface than the stated 1.6 cm. A 0.5 mm change in the SSD would yield a 6 % difference in dose due to the inverse square law and would improve the agreement between the measured values and the TG43-EQ profiles. Another hypothesis is that the difference could be due to the large measurement errors due to the uncertainty in the GafChromic film’s dose response. Figure 4.6 shows two measurements for the H10 applicator at 3.38 mm depth supporting this theory. The mean difference between the TG43-EQ and the measured profile decreased from 11.2 % to 5.1 % when remeasured with HS film, which has a lower uncertainty than the MD55 model. These are however just hypothesis and further measurements should be done, preferably with higher precision, in order to make an appropriate conclusion. From the measurements done it is impossible to say whether the difference is due to the uncertainty in the GafChromic film’s dose response, due to a misplacement of the radioactive source or due to a combination of both.
Figure 4.6: The TG43-EQ profile (perpendicular to the source) was compared with two measured profiles for the same applicator and at the same depth. Using the HS film model (the lower profile) decreased the mean difference from 11.2% to 5.1% in this case. The TG43-EQ model is still predicting a continuously lower dose rate throughout the entire profile, which could be explained with a small displacement in the source position. However, more measurements should be done before a conclusion can be made.

4.2 The asymmetric dose rate profiles for the vertical applicators

Figure 4.2 in section 4.1 showed that the measured profiles for the vertical applicators had a non-symmetrical shape relative to the central axis. The difference between the region with higher dose rate and the region with lower dose rate for these particular profiles was in the order of 9% for the V10 applicator, 12% for the V20 applicator and 15% for the V30 applicator. This kind of asymmetry was observed for all the measurements that were done with the vertical applicators, even though the experimental set-up was changed in order to try to get a more symmetric profile. The table below shows the measured difference between the area with higher dose rate and the area with lower dose rate for some profiles that were measured on different occasions. The results show that the asymmetry is not always the same even though the same applicator is used.

<table>
<thead>
<tr>
<th>Applicator</th>
<th>Area with lower dose rate [cGy/h*U]</th>
<th>Area with higher dose rate [cGy/h*U]</th>
<th>Difference %</th>
</tr>
</thead>
<tbody>
<tr>
<td>V10</td>
<td>0.27</td>
<td>0.32</td>
<td>19</td>
</tr>
<tr>
<td>V10</td>
<td>0.275</td>
<td>0.3</td>
<td>9</td>
</tr>
<tr>
<td>V20</td>
<td>0.27</td>
<td>0.31</td>
<td>15</td>
</tr>
<tr>
<td>V20</td>
<td>0.25</td>
<td>0.28</td>
<td>12</td>
</tr>
</tbody>
</table>

Table 4.1: The measured differences between the areas with higher dose rate and the areas with lower dose rate for the V10 and V20 applicators. The measurements were done on two different occasions.
A possible explanation to the asymmetric profiles is that there might be a small tilt in the source position meaning that the incoming source does not lie perfectly parallel with the central axis but rather with a certain tilt angle, $\theta_{\text{tilt}}$. To control if this assumption is correct and to find out the tilting angle, the TG43-EQ program was modified in order to simulate the incoming $^{192}$Ir source tilting at a certain angle. Fig 4.7 shows the tilting angle model compared with the measured model for the vertical applicators.

Figure 4.7: The TG4-EQ3 model was modified to simulate a tilting source relative to the central axis for the vertical applicators. The profiles have been scaled in order to be able to compare the shapes of the TG43-EQ tilting model and the measured data. The tilting angle is 4 degrees for a) the V20 and b) the V10 applicators and 3 degrees for the c) V30 applicator.
Both profiles were normalised to the dose rate at central axis before they were compared, in order to get a better view of how the forms between the tilting source model and the measured data agrees. The best fit between the two models was found using a tilting angle of 4 degrees for the V10 and V20 applicators and 3 degrees for the V30 applicator.

Using the vertical applicators for treatment without knowledge of this possible tilting angle could compromise the outcome of the treatment since the difference in dose between the high dose regions and the low dose region is in the order of 15 % for the V10 applicator, 7% for the V20 applicator and 13% for the V30 applicator. Furthermore table 4.1 shows that the difference between the high and low dose regions is not always the same for the same applicator making it difficult to correct for the tilting source.

The asymmetric dose distribution will have a large impact on the dose received in the penumbra region especially at depth. Since the dose will be significantly higher on one side than on the other, the dose fall-off at the edges of the treatment area will start falling off at different distances from the central axis. As an example consider a tumour being treated with the V20 applicator with a certain tilt angle in the source position as to receive the measured profile in figure 4.8. In this particular case a point located at 1.4 cm from the central axis will receive a 9 times higher dose if it is located on the high dose rate side than if located on the low dose rate side. Since it is normally healthy tissue that is located at the edges of the treatment area this effect must be taken into consideration to avoid inducing unnecessary damage to healthy tissue.

![Figure 4.8](image)

**Figure 4.8:** The dose rate distribution for the V20 applicator at 3.38mm depth with a certain tilt angle in the source position. The figure shows the large differences in the dose received by points in the penumbra regions.
Another problem is the change in the local minimum dose at the central axis. The local minimum dose will no longer be located at the central axis but be skewed to one side depending on the tilt angle. This can cause problems since the prescription dose is prescribed to the central axis. Prescribing the dose to the central axis will lead to that there will be parts of the tumour receiving lower doses than the prescription dose. In order to investigate what impact this difference in dose might have on the treatment outcome, the biological effective dose, BED, for a normal treatment scheme is examined. The BED is given by:

$$BED = nd \left(1 + \frac{d}{\alpha/\beta}\right)$$

where $n$ is the number of fractions
$d$ is the dose given per fraction
$\alpha$ and $\beta$ are constants characteristic for a certain type of tumour or tissue.

The number of fractions for a normal treatment with the Leipzig applicators is 6 and the prescribed dose to the central axis is 6 Gy per fraction. The constants $\alpha$ and $\beta$ are not known for the specific tumour but whatever the values of $\alpha$ and $\beta$, the ratio $\alpha/\beta$ for malignant tumours is assumed to equal 10.

The BED is a method of quantifying the effect of a fractionation scheme. Giving a total dose of for example 36 Gy in 6 fractions will have a different effect on the tumour and on the normal tissue than giving 36 Gy at once or in 3 fractions. The figure below shows the BED for the tilting source dose profiles from Fig 4.2.
The BED at central axis for a normal treatment scheme with 6 fractions and 6 Gy per fraction is 57.6 Gy. If the tumour is not larger than the cone diameter, prescribing this dose to the central axis results in doses ranging from 55.8 Gy at the real minimum point close to the central axis up to 66 Gy in the high dose region on one side of the central axis for the V10 applicator. Similar dose intervals are obtained for the other applicators. When calculating the BED to the tumour an assumption has been made that the applicator is positioned at exactly the same position every time, which of course is not the case for real treatments since it is very difficult to place the applicator in exactly the same way. Due to lack of time, the effect of misplacements of the cone has not been investigated.

The tilting source presents another difficulty as well, namely to which point should the dose be prescribed? Depending on the tilt angle of the source, the position and the magnitude of the minimum dose will vary. Not knowing where the actual minimum point of the dose distribution will occur and how much lower the dose will be, might make it troublesome since it will be hard to tell if the entire volume of the tumour will get enough dose in order to have sufficient tumour control probability.
The dose volume histogram for the tumour will thus change with the tilting angle. To investigate how it changes, a hypothetical cylindrically shaped tumour is assumed to be located one mm below the skin surface. The size and shape of the tumour is shown in the figure below.

Due to its size the hypothetical tumour is treated with the V20 applicator. The TG43-EQ dose profiles at 0.1 and 0.3 cm depth are used to approximate the absorbed dose to the tumour. A logarithmic interpolation is used to obtain the absorbed doses in between these depths. The figure below shows the change in dose volume histogram due to the 4 degree tilt in the source position.
The dose volume histograms for the virtual tumour show that larger volume of the tumour receives higher dose when using a non-tilting source than when using a tilting source. If the hypothetical tumour is located in the same position as mentioned above, the minimum BED received by the entire tumour will be 52.5 Gy if using the tilting source and 57.6 Gy if using the non-tilting source, thus a 9% difference in radiobiological effective dose.

Using the tilting source during an entire treatment of 6 fractions and 6 Gy per fraction will thus result in a decrease of the BED of approximately 9%.

This implies that overdosing a small area by 15% does not compensate for the loss of tumour control probability in the low dose area.

These differences in the dose and the tilting angle occurred even though an experimental set-up was used to try to avoid the source from tilting. In real life the skin applicators are positioned in all sorts of positions, which probably will result in larger tilt angles and thus larger variations in the dose distributions. Therefore one should be really careful and be aware of this particular problem with the vertical Leipzig skin applicators when they are used in clinical practice.
5 Conclusions

Theoretical modelling of dose distribution under the Leipzig applicators using an analytic approach was very challenging. Significant and interesting differences between theory and experimental results were noted.

A conventional flat-bed scanner was found to be acceptable for use in order to read out the measured absorbed dose of GafChromic film pieces with uncertainties of approximately 1.2 %. Together with the uncertainties in the film’s response the total error in the measured results is approximately 5%. The scanner accuracy is dependent on the scanning geometry, the position dependence, the linearity of response and the uniformity of the scanner and should be checked before using the scanner for measurements.

The TG43, our modified TG43-EQ and the Monte Carlo theoretical curves were proved to be generally similar, except for differences due to scattered radiation from the collimator walls surrounding the $^{192}$Ir source and to the differences in the full scatter conditions near the treatment surface. Therefore, the most crucial influence on dose distribution is the geometric position of the source above the skin surface. The mean difference between the TG43 and the Monte Carlo models for nine different profiles compared at depths ranging from 0.1 - 0.5 cm is 2.11 % for the vertical applicators and 1.25 % for the horizontal applicators. Using the TG43-EQ model, the mean difference for the same profiles at the same depths decreases to 1.75 % for vertical applicators and 1.22 % for the horizontal applicators. Taking the scattered radiation from the collimator walls into account would probably improve the agreement even more.

More alarmingly, a significant discrepancy was noted between the theoretical distributions (assuming perfect set-up) and real experiments that cannot be explained by random uncertainties in dose measurement. One aspect was consistent with a tilt in the vertical source position leading to asymmetric profiles. The tilt in the source position was shown to affect the dose distribution, leading to an underdosage of 6 % in the BED to the tumour. This difference was present even though a special set-up was used to try to get the source coming in vertically. In practise the underdosage might be even larger since the applicators are placed in all kinds of positions, possibly increasing the tilt in the source.
A couple of profiles showing another possible displacement in the source position leading to a decrease in the SSD, which in turn leads to a higher than predicted dose to the tumour, were also measured. This however, is more acceptable in terms of tumour control than underdosed volumes. Time did not permit us to explore modifications of our TG43-EQ model to account for variations in the SSD. In order to make a conclusion for this case, more measurements, preferably with better precision, are needed.
References


ICRU 38, Dose and Volume Specification for Reporting Intracavitary Therapy in Gynaecology, International Commission on Radiation Units and Measurements (ICRU 38, Bethesda, MD 1985).

ICRU 60, Fundamental Quantities and Units for Ionising Radiation, International Commission on Radiation Units and Measurements (ICRU 60, Bethesda, MD 1998).


www.ispcorp.com/products/dosimetry → products → GAFCHROMIC