Radiation dosimetry in the Leksell Gamma Knife®
treatment room

Experimental measurements vs. Monte Carlo simulations
Abstract

The medical technology company Elekta is the manufacturer of the worldwide used Leksell Gamma Knife® (LGK), an instrument specially designed for treating brain disorders. Sweden today has two LGKs treating patients, one of an older model and one newer. Both are owned by the Karolinska Hospital and placed in Stockholm. The radioactive sources of Cobalt in the machine lead to an enhanced radiation level in the treatment room and it is therefore important to have sufficient shielding in the surrounding walls.

In this Master of Science Thesis the radiation in the LGK treatment room of both these machines has been measured, using a sodium iodide scintillator detector. The absorbed doses are calculated by integrating the resulting energy spectra from the detector-registered particles.

A Monte Carlo simulation program is used to simulate photon transport through the LGK machine. The absorbed doses and detector count rates are calculated and compared to the experimental values.

The results show the differences between the photon transport in the two models, and that the radiation leakage is much higher in the older model. The comparison between the simulated results and the experimental results shows excellent agreement on the backside of the machine. This suggests a simulation method applicable for particle transport through the LGK.
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1. Introduction

1.1 Background

About 1000 people are diagnosed with brain tumour in Sweden every year. Over the last decades stereotactic radiosurgery has become more and more important in the treatment of brain tumours and it is a favorable option to conventional surgery, especially when the tumour is located deep within or close to vital areas. [8]

The technique is based on using multiple beams of radiation, focusing on a small volume. This way a high dose can be delivered to the target, while keeping the dose in the surrounding tissue low. Very commonly used as radiation is X-rays, produced in linear accelerators. In the Leksell Gamma Knife® (LGK) a machine specially designed for treating brain disorders with stereotactic radiosurgery, gamma radiation from radioactive sources is used.

The LGK, which is manufactured by Elekta, was first developed for treating brain tumours but today it has a widespread use in clinical medicine, such as blood vessel defects, Parkinson’s disease, epilepsy and obsessive compulsive disorders (OCD). Approximately 200 000 patients have been treated with the Gamma Knife and over 200 machines have been exported to all over the world. [8]

Elekta, which is an international, stock listed company in medical technology, was the first to develop stereotactic radiosurgery and their main product is the LGK. The company was founded in 1972 by Lars Leksell, and has since then been working with developing clinical solutions and systems for radiation treatment of brain disorders. In addition to Gamma Knife surgery and linac based stereotactic surgery Elekta also develops other products in areas such as functional mapping, image guided radiation therapy, stereotactic radiation therapy etc. [14]

The name Leksell Gamma Knife® doesn’t refer to a real knife performing a surgery, but to the knifelike precision and power of the collimated photon beams that are used to irradiate the target area. The gamma photons come from radioactive sources of Cobalt that are placed in pellets inside a shielding ball of steel in the machine. Therefore the gamma radiation reaching outside the machine will to large extent be reduced in intensity and in energy, mostly due to Compton scattering. (See further in chapter 3.1)

The shielding in the walls of the Gamma Knife room must be designed so that the personnel working with and around the machine don’t receive higher doses than allowed. The maximum allowed doses from ionizing radiation in Sweden is determined by the Swedish Radiation Safety Authority (SSI). For a person working with radioactive materials, e.g. in a hospital, the maximum effective dose allowed is 50 mSv during one year, but only 100 mSv during any period of five years. [13]

Today the following guidelines are used to give satisfactory radiation protection, given in thickness of concrete wall (the values are taken from measurements performed on different Gamma Knifes around the world);
Wall in front of the LGK  600 mm concrete
Floor, ceiling and walls on the side  500 mm concrete
Wall behind the LGK     300 mm concrete

However, if possible, it is desirable to reduce the shielding since it would lead to a lower installation cost for the buyer.

1.2 Aim of the work

The aim of this project is to map the radiation dose distribution in the LGK treatment room. A scintillator detector is used to measure the energy spectrum of the gamma radiation in different positions around the machine.

The gamma photons, travelling through the shielding material, will either be absorbed or reduced in energy due to Compton scattering. Therefore only a small part of the primary full-energy photons will be shown in an energy spectrum collected. By integrating the energy spectrum the absorbed dose in the detector can be calculated.

Monte Carlo simulation of the dose distribution will be performed on a model of the machine and used to compare with the measured results. Hopefully it will lead to verification that these kinds of simulations can be performed for radiation transport in the LGK.

Further, if the simulation and measurements are in agreement, this simulation technique may in the future be used to determine the shielding dimensions of the LGK treatment room for each specific case.
2. Electromagnetic radiation

2.1 Interactions in matter

2.1.1 Gamma radiation

To understand the nature of the LGK technique it is necessary to first look at the basic characteristic of gamma photons and effects of ionizing radiation in living tissue.

Electromagnetic radiation can be described as a flow of photons, which are uncharged and massless particles, all traveling with the speed of light. Each photon contains a certain amount of energy. Photons that lie in the same energy range define the various types of electromagnetic radiation. Radio waves and microwaves are low energy photons, while infrared, visible and ultraviolet light have more energy. Most energetic are X-rays and gamma rays.

Both gamma radiation and X-rays are examples of electromagnetic radiation that are useful in cancer treatment, since they are high-energy photons and relatively easy to produce. They both have the same properties but differ in the way they are produced. (Generally all types of electromagnetic radiation have the same properties, but low frequency photons tend to have more wave-like behavior while for very high frequency photons particle-like properties dominate. The types may also differ in the way they are produced and can be detected.) [2]

The gamma photons might, like alpha and beta particles, be emitted from atomic nuclei in radionuclide decays. All radioactive nuclei will emit radiation with one or several energies, specific for the isotope.

X-rays used for medical treatment, also referred to as Bremsstrahlung, can be produced for example in linear accelerators (linacs). A high voltage is used to accelerating electrons and the photons are emitted when the electrons are decelerated upon collision with a metal target. Linac produced X-ray normally have energies $\geq 4$ MeV. X-rays can also be emitted when excited atoms are de-excited; electrons from higher states drop down to fill the vacancy, emitting a photon with energy corresponding to energy difference between the electron levels. These are called characteristic X-rays. [3]
Photons lose their energy by interacting with the media they are passing through. There are three main interactions for electromagnetic radiation; photoelectric absorption, Compton scattering and pair production. The probability, for each interaction, normally given as the interaction cross-section, depends on the energy of the photon and the atomic number of the interacting media. [3]

2.1.2 Photoelectric absorption

In photoelectric absorption the incident photon deposits all of its energy to an atom (and thereby ceases to exist). The atom moves to an excited state and a photoelectron is released from one of the shells, with energy equal to the photon energy minus the binding energy (b.e.) of the electron. It is therefore necessary for this process that the energy of the photon exceeds the binding energy of the electron.

Since the loss of an electron leads to a vacancy in one of the inner shells, the atom sometimes also emits fluorescent radiation. This radiation is released when the vacancy is filled by electrons from outer shells or by capture of a free electron.

The probability for photoelectric absorption will increase with the atomic number, \(Z\), of the absorber (\(\sim Z^4\)), and will decrease as the energy of the photons increases. This type of interaction is therefore only dominant at very low energies.

2.1.3 Compton scattering

Compton scattering (coherent scattering) is more probable at higher energies. In this process a photon loses part of its energy to an atomic electron through collision. After the collision the out-going photon will be scattered in a new direction, \(\theta\), relative to the original photon (see figure 2.2). The photon lose energy and the probabilities for new interactions are changed.
With the geometry according to the figure above and the requirement of energy and momentu

\[ \Delta E = \frac{hc}{\lambda_f - \lambda_i} \]  

\[ \Delta \lambda = \lambda_f - \lambda_i = \frac{h}{m_0 c} (1 - \cos \theta) \]  

2.1.4 Pair production

Pair production takes place inside the coulomb field of the nucleus. In this process the atom absorbs the incoming photon and an electron-positron pair is created. Due to the creation of new particles, this process is only possible for photon energies above 1.022 MeV, corresponding to the rest mass, \(2m_e c^2\), of the electron-positron pair. Since the energy and momentum must be conserved also in this process all energy that exceeds 1.022 MeV becomes kinetic energy, divided between the two new particles.
After a pair production, when the positron is slowed down, two annihilation photons are produced.

The total probability for a photon to undergo interaction can be plotted as a sum of the partial probabilities for photoelectric absorption, Compton scattering and pair production;

\[ \mu = \mu_{\text{photo}} + \mu_{\text{compton}} + \mu_{\text{pair}} \]  

(eq. 2.3)

![Figure 2.5](http://nuclear.ph.man.ac.uk/~jb/n03/NE03(2).ppt, December 2005)

At energies corresponding to the binding energy of an atomic shell there are discontinuous jumps in the probability function for photoelectric absorption. These are so-called absorption edges. When the binding energy of the K-shell is approached from higher energies the probability increases until the point where the energy is no longer sufficient to release the K-electrons and the function drastically drops (see figure 2.4).

There are also other types of possible interactions for photons. For example, a photon may also scatter on an atomic electron and head off in a new direction, but without losing any of its energy, so called Rayleigh scattering or coherent scattering.

### 2.2 Exponential attenuation

Unlike a charged particle, that always loses some or all of its energy in a collision, gamma radiation has no finite range. The photons will travel until all of their energy has been deposited through interaction. As stated above, the probability for each type of interaction is depending on the energy of the photons and the atomic number of the
material they are passing through. There is also a probability for a photon to travel through a media without undergoing any interactions. If we consider a beam of monoenergetic photons passing through matter of thickness, $x$, the intensity of the beam will be attenuated for each photon that loses energy and/or direction. The loss of intensity in the beam, $I(x)$, can be described as an exponential decreasing function that varies with the thickness of the absorbing material,

$$ I(x) = I_0 e^{-\mu x} \quad \text{(eq. 2.4)} $$

where $I_0$ is the intensity of the initial beam and $\mu$ is the total linear attenuation coefficient, given in length$^{-1}$ and is also defined as the inverse of the mean free path of the particle, $\lambda^{-1}$.

As stated above the total linear attenuation coefficient will be the sum of the probabilities for the different types of interactions, $\mu = \mu_{ph} + \mu_c + \mu_{pp}$, and can be seen as the total probability for losing a photon from a collimated monoenergetic beam. Since the linear attenuation coefficient is dependent on the density of the absorbing material it will be different depending on the physical state of the material. More commonly used is the mass attenuation coefficient, $\mu/\rho$, where the coefficient is being normalized by dividing with the density of the material.

![Figure 2.6 Narrow beam attenuation.](From Turner, J.E. *Atoms, Radiation, and Radiation Protection*, 2nd ed. New York: Wiley-Interscience, 1995 Fig. 8.7)

In figure 2.6 only the primary photons hit the detector and it is referred to as narrow-beam geometry, where the secondary and scattered particles are excluded. The linear attenuation coefficient, $\mu$, also describes a narrow-beam attenuation since it only includes the primary particles.

In practice, a narrow-beam geometry can be difficult to achieve since the detector must be placed so far away or in such position that every scattered or secondary particle with any deflection miss the detector. Another way of receiving narrow-beam attenuation is to only count the registered particles with the full energy, right direction and photon characteristics, such as the speed of light.
If scattered and secondary particles also are included, the value of the attenuation coefficient, \( \mu \), will decrease, and the attenuation is referred to as broad-beam. The factor between narrow-beam attenuation and broad-beam attenuation is referred to as the build-up factor.

### 2.3 Absorbed dose, linear energy transfer and quality factor

The energy lost by photons (and other radiation particles) is absorbed by the media since the energy is transferred to the secondary particles and the atoms and molecules in the media. We define the absorbed dose, \( D \), as the absorbed energy per unit mass. The SI unit is Gray (Gy), where \( 1 \text{Gy} = 1 \text{ Joule/kg} \).

The energy transferred from a particle, per unit length of the ionizing track, is defined as the linear energy transfer (LET). The unit is given in keV/\( \mu \text{m} \). In radiobiology particles are often categorized in high-LET and low-LET particles, and can be used as a reference to the effect that different types of radiation have on living tissue. Heavy ions and alpha particles are considered high-LET particles, while photons are low-LET.

Due to the linear energy transfer the biological effect of the radiation will not only depend on the energy of the particles but also on the type of the particles. For example the energy from an alpha particle can be 20 times more effective on living tissue than the same amount of energy from a gamma photon. The absorbed dose only tells the amount of energy, but if this value is multiplied with the effectiveness or the quality factor, \( Q \), of the particle one obtains the dose equivalent, \( H \);

\[
H = D \cdot Q
\]

which gives the biological effect. Since the quality factor is dimensionless the unit dimensions joule/kg will be unchanged, but applied to the dose equivalent the unit is called Sievert. The quality factor of a particle is determined of the linear energy transfer value.

### 2.4 Cancer treatment with radiation

There are various ways of treating cancer. The most commonly used are; chemotherapy, surgery and radiation treatment and they are often used in combination with each other. Radiation treatment can for example be used before surgery to “shrink” the tumor or after surgery to irradiate cells infiltrating surrounding tissue. [8]

In surgery the tumour is completely or partly removed from the patient. This form of treatment is only possible if the cancer isn’t spread over larger areas in the body.

Chemotherapentic drugs is a group of drugs developed to interfere with the division of the cancer cells, and thereby keep them from growing. They are focused to only work on cells that tend to multiply rapidly, and it therefore makes it possible to distinguish the cancer cells from the healthy ones. The biggest side effects from chemotherapy are the
killing of normal cells that also divide themselves rapidly and the chance of that the cells develop resistance to the drug. [10]

Radiation treatment is preformed both internally (so called brachy therapy) and externally. In both cases the tumour is exposed to ionizing radiation, where the energy of the radiation is sufficient to remove an electron from one of the atomic shells or disrupt the bonds between atoms and molecules.

In living tissue the ionizing radiation can damage the cells. The ionized molecules and free electrons, moving in the body, can react with other atoms and molecules in their way, causing free radicals and biochemical damages on the DNA.

The gene pool of a molecule consists of two equal chains of DNA. If one of the chains is damaged, a so-called single strand break, the cell is able to repair the chain, with the other one as reference. But if both of the chains are damaged, so-called double strand break, the structure of the DNA will be destroyed, and the cell loses its ability to reproduce itself.

There are two main groups of ionizing radiation that are used for treating cancer; electromagnetic radiation, such as x-rays and gamma radiation, and particle radiation, for example protons.

Particle radiation has mass and the particles are decelerated as they move through a media. Therefore they will deliver most of their energy in the end of their range and before they stop there will be a top in the energy loss curve, the so-called Bragg peak. This peak is applied in radiation therapy, to reach deep laying targets with high doses, without damaging the surrounding tissue. Particle beams are produced and accelerated in particle accelerators, such as linacs, synchrotrons and cyclotrons.

Electromagnetic radiation, with no mass, travels with the speed of light all the time, until all energy has been deposited through interaction. Thus, the higher energy, the higher probability the photons will travel further. Radioactive sources are used to generate gamma radiation while medical X-rays normally are produced in linear accelerators, but there is no difference in how they affect the living cells.

Figure 2.7 DNA base damage, DNA single strand break and DNA double strand break.
In radiation therapy only small doses are delivered at a time and the treatment is repeated over a period, so called fractional radiation therapy. The irradiated healthy cells are therefore able to self-repair between the irradiations, while cancer cells are supposed to have more difficulty to repair themselves than normal cells.

When treating brain tumors the spatial margins are often small and it is important that the normal brain tissue surrounding the tumor is protected and receives as little amount of radiation as possible. Stereotactic radiosurgery, SRS, are often used in this case as an alternative to the above described treatment options. The idea behind the technique is to use multiple beams of radiation, converging in three dimensions, to focus on a small volume. This way a high dose is delivered to the target volume, while keeping the dose in the surrounding tissue low. The high precision is achieved by head fixation and accuracy of the radiation field. Stereotactic radiosurgery is preferably performed with the Leksell Gamma Knife®, due to the greater accuracy of target positioning compared to linac-based stereotactic radiosurgery.
3. Stereotactic radiosurgery with the Leksell Gamma Knife®

3.1 The Leksell Gamma Knife®

3.1.1 Basic idea and design

The Leksell Gamma Knife®, manufactured by Elekta Instrument AB, is designed exclusively for treating brain disorders using the technique of stereotactic radiosurgery. It uses multiple beams of gamma radiation to focus precisely on the lesion. The idea is based on the “center of arc” principle, where the target volume is at the center of a circular arc of rotation. Thus, the target can be approached in three dimensions. The individual gamma rays are too weak to significantly harm the cancer cells in the tumor, but when multiple beams are focused to the same point the target will receive the sum of radiation from the different beams while the surrounding area only receive a minimal dose.

In contrast to radiation therapy, when the irradiation is repeated, stereotactic radiosurgery delivers a single high dose to the target volume, and thereof the name surgery.


The gamma radiation is emitted from 201 sources of $^{60}$Co, distributed along parallel circles. The photons, emitted from each source, are collimated into beams and focused to a common volume, with a source to focus distance of 400 mm.

There are two types of collimators installed in the machine to focus the radiation but the beam diameter is defined by the size of the third and final collimator, placed in an
interchangeable helmet. There are four different helmets with collimator sizes; 4, 8, 14, 18 mm. One of the helmets will be inserted in the machine during treatment. The patient’s head is attached with screws to an aluminium head frame, connected to the collimator helmet. The head frame is remotely adjustable for all three directions. The values of the coordinates are determined in the planning process and the frame can be attached so that the target is placed exactly in the focal point of the collimated beams.

The sources are placed inside a large shielding ball of steel, so that the gamma radiation will be absorbed or energetically degraded through Compton scattering before it exits the machine.

As mentioned above, to avoid damaging the surrounding healthy tissue this technique requires high spatial resolution, since the size of the target volume are often very small. The ability to perform stereotactic radiosurgery decreases with the increasing number of targets, and is not suitable for targets larger than a few centimeters.
3.1.2 The sources

The gamma radiation is emitted from sources of \(^{60}\text{Co}\). Cobalt is a transition metal with many isotopes, and the majority of them are unstable. \(^{60}\text{Co}\) is produced from the naturally occurring isotope \(^{59}\text{Co}\) by adding an extra neutron to the nucleus (done by bombarding \(^{59}\text{Co}\) atoms with neutrons). The \(^{60}\text{Co}\) atom decays to \(^{60}\text{Ni}\) by emitting a beta particle, with energy 0.32 MeV, and two gamma photons, with energies 1.17 MeV and 1.33 MeV. Since the energy of the beta particle is so low it will be absorbed before reaching outside the source capsules. [17, 18]

One of the reasons for choosing cobalt-60 is its suitable half-life of 5.27 years, but also the high energy of the gamma photons emitted. Thus, the dose rate will decrease with time and the irradiation time during the treatment must increase. The machine is loaded with new sources with approximately 5 years interval. The loading procedure is a rather complicated process and may take several days to complete.

3.2 Treatment

3.2.1 Treating area

The LGK is used treating a number of disorders within the brain; primary malignant brain tumours (that arise in the brain) or metastatic malignant tumours (that arise in some other part of the body but spread to the brain), benign brain tumours, arteriovenous malformations (blood vessel defects) and functional problems such as trigeminal neuralgia (facial pain). Treatment for patients with Epilepsy and Parkinson's disease are evaluated for the future and recently the LGK also has been used in so-called psychosurgery e.g. for treatment of obsessive compulsive disorders (OCD).

The abnormal cells in a brain tumour can directly destroy the healthy normal cells, but they can also cause damage in form of inflammation, pressure and swelling. [8]

Compared to conventional open surgery the risks of complications are relatively low and the experience is probably less inconvenient for the patient. Sometimes open surgery is not even an option, because of the patient's health or age.

3.2.2 Treatment procedure

The LGK procedure is a single session treatment and it is often completed in one day. Normally the patient can go home the same day and continue with normal activities.

The treatment session involves four different steps. First is the attachment of the head frame, where the patient is set under local anesthesia and the head frame is fixed with screws to keep the head completely still during imaging and treatment.
Second the patient has to undergo imaging studies. CT-scans (Computed Tomography) or MRI-scans (Magnetic Resonance Imaging) are used for imaging of the tumour and to show the exact location in relation to the head frame. For visualizing brain tumors and trigeminal nerves MRI is preferable, but CT can be used e.g. when the patient has a pacemaker and it is not suitable to perform a MRI examination.

In the third step the images are used in the computerized dose planning program to calculate how an optimal dose can be delivered, which also includes the length of the treatment. The targets are often best treated by combinations of several so-called shots delivered to different parts of the tumour. The software program also includes simulation of the spatial distribution of the dose.
The last step is the irradiation procedure. After the dose planning is completed the patient is placed on the treatment bed and the head frame is connected to the collimator helmet. When the doors on the machine open the bed is moved backwards so that the collimator helmet can be locked to the collimators inside the machine.

The operating team normally consists of a radiation oncologist, a medical radiation physicist and a neurosurgeon, neurologist and a radiation therapy nurse. The patient can’t
feel the radiation. The team can’t stay in the room because of the high radiation level when the shielding doors are opened and the maneuvering is performed remotely. A camera is placed in front of the LGK so that the operating team can follow the course of events, and the patient is able to communicate with them through a microphone in the helmet.

Depending on the shape and size of the target and the dose required, the length of the treatment varies from minutes up to a few hours. Following the treatment the tumours shrink over a period, malignant and metastatic tumours shrink within a couple of months while benign tumours can take up to two years.

3.3 Technical development and usage in Sweden

The Gamma Knife was invented by Dr. Lars Leksell, who also was the first to perform stereotactic radiosurgery. This was done at the University of Uppsala, where the operation could be performed with a cyclotron. Due to the lack of an accelerator Dr. Lars Leksell treated the first Gamma Knife patient in 1968 at the Karolinska Institute in Stockholm using radioactive sources.

The first version of the Gamma Knife was the Model U, followed by the slightly different Model B. The machine installed at Radiumhemmet today is of the Model C. It has the same accuracy as the two earlier models, U and B, but with the addition of the Automatic Positioning System. With this system the machine automatically can change the position between the “shots”, which earlier had to be made manually by the staff. This leads to a reduced treatment time for the patient.

Sweden has two LGKs treating patients today, one at Radiumhemmet, Karolinska Hospital and one in Sophiahemmet hospital in Stockholm (the latter recently bought by Karolinska Hospital). KS are treating about 320 patients per year. [8]

The cost of a LGK machine is about 3.5 million dollars, which can be seen as a rather large investment for a hospital. But on the other hand the surgery with LGK is often 25 to 30 percent cheaper than traditional neurosurgery and since the patient can go home the same day the hospital isn’t burdened with long hospitalizations. [12]
Figure 3.5 The Gamma Knife installed in Radiumhemmet, at the Karolinska hospital in Stockholm, together with the detector system used in this project.
4. Measurements of dose distribution around the Gamma Knife

4.1 Method and equipment

4.1.1 The detector and electronics

Gamma photons can, since they are neutral particles, only be detected indirectly, through the secondary particles from interactions. One of the oldest techniques of detecting ionizing radiation is through scintillation, where the energy lost by ionization is converted into pulses of light by a scintillating material. [3]

The measurements of the dose distribution around the LGK were performed with a scintillating NaI-detector (sodium iodide), which consists of a NaI-crystal (2*2 inches cylinder) connected to a photomultiplier tube (PM-tube). The detector is covered by a thin reflecting material and a 0.5 mm aluminium shield.

Sodium iodide is an inorganic scintillating material commonly used in radiation detectors, mainly because of its high scintillation light output and economical advantages. Inorganic materials are often doped with an impurity to make the light conversion more efficient. This impurity is called an activator and in NaI-detectors TI (thallium iodide) is often used. [3]

The photons enter the detector through a front window and interact with the atoms in the crystal. The photon energy is transferred in the crystal to the electrons and positrons from the interaction processes. These secondary particles will then, with gained kinetic energy, ionize and excite atoms in their way, causing the crystal to emit small flashes of light (scintillate). Since the mean free path of the electrons is of the order of µm it is assumed that all of their energy will be absorbed in the detector.

The scintillation light is emitted when the atoms are de-excited and the amount will be proportional to the kinetic energy of the electrons. The conversion property of the detector, ionization to light, increases with higher density and atomic number, Z of the crystal.

Both organic (e.g. liquid and plastic) and inorganic (alkali halide crystals) materials are used in scintillators, giving different properties of the detector. [3] For example, plastics are used in timing applications since they have much faster pulse light decay, while the energy resolution is very bad.

<table>
<thead>
<tr>
<th>Scintillator</th>
<th>Density (g cm(^{-3}))</th>
<th>Effective Z</th>
<th>The total light yield (photons/MeV)</th>
<th>Decay constant (ns)</th>
<th>Wavelength of emission (nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sodium Iodide (NaI)</td>
<td>3.67</td>
<td>50</td>
<td>38000</td>
<td>230</td>
<td>415</td>
</tr>
</tbody>
</table>

**Table 4.1** Characteristics of the used NaI-crystal
Next the visible photons enter the PM-tube, which will convert the light into electric pulses, giving an output signal proportional to the deposited energy. The light first strikes the cathode, a photosensitive surface that can release electrons through the photoelectric effect. The probability that an electron is emitted from the surface is referred to as the quantum efficiency of the photocathode.

A voltage is applied to accelerate and focus the released electrons between dynodes in the tube. When an electron hits a dynode secondary electrons are emitted from the surface. For each dynode in the chain more and more electrons are released and the photoelectron signal can be gained up to $10^7$ times through the PM-tube. All the electrons are collected at the anode.

The electric signal from the detector is sent to an amplifier, which is used to amplify and shape the electric pulses. The output signal is sent to a Multi Channel Analyzer (MCA) with a computerized read-out so that the energy spectrum can be viewed on a computer.

The energy resolution of the detector is normally given as the width of an energy peak at half maximum (FWHM), describing the detector ability to discriminate between photons with almost the same energy. The energy resolution is a function of photon energy, but also depends on the size and material of the detector. A typical value for a small NaI detector for 662 keV photons is about 7%, for lower energies the resolution is worse, e.g. 45 % for 5.9keV photons.

4.1.2 Performance

The measurements of the dose distribution were performed on the Gamma Knife at Radiumhemmet (Karolinska Hospital in Stockholm) and at Sophiahemmet Hospital in Stockholm.

The Gamma Knife at Radiumhemmet was installed in 1988 and is one of the oldest models. From the beginning it was the prototype for the Model B and has later been upgraded with the Automatic Positioning System and extra shielding.

The Gamma Knife at Sophiahemmet is from the beginning a Model C and therefore newer. It was installed in 2001.
As will be seen in the results the radiation “leakage” is larger in the older model. There are differences in the lead shield in the front and in the thickness of the shielding, which is smaller on the older model, leading to more photons with higher energy reach outside the machine.

The measurements with the scintillator were first performed on the Gamma Knife at Radiumhemmet. Since the Monte Carlo simulations (see next chapter), which are based on a Model C, differed several orders from the results the same measurements were also performed at Sophiahemmet (where the Gamma Knife corresponds to the model used in the simulations).

First a mapping of the room was made where the points for the measurement were selected. (See figure 4.2) The center of the shielding doors was chosen as the center point.

The detector was mounted on a support, connected to a cart, thus the height could be adjusted and the detector could easily be moved between the measuring points and be kept in position during collecting the data.

To get a dose distribution in three dimensions the measurements were performed in three different plans; one near the floor, one at the level of the radiation focus and one above the machine.

The threshold energy in the measurement was set between 25-30 keV.

Symmetry around the x-axis was assumed and the measurements were therefore only performed on one side of the machine (see figure 4.2).

![Figure 4.2 Map over the measuring points in the gamma knife room at Radiumhemmet.](image)

Data was collected for 100s in each point.
In some places near the machine and when the shielding doors are open the acquisition rate is too high for measuring the radiation with the NaI-detector. Another much smaller scintillating material would be needed, with higher density and atomic number, Z, to maintain the photopeak to Compton distribution ratio. Therefore some of the points in front of the Gamma Knife are omitted (see figure 4.2).

To verify earlier dose distribution in the therapy room, given in μSv/h, measurements were performed simultaneously with the scintillator using two dosimeters; a Smart-ION from Mini Instruments and an intensimeter, RNI-10. The findings were in accordance with the earlier measurements. The deviations detected were interpreted due to different cut-off at low energies.

4.1.3 Background in the spectra

Due to the cosmic radiation and the existence of natural radioactivity in our surroundings (e.g. in the ground and in construction materials) the spectra also include background radiation. [3]
At Radiumhemmet the Gamma Knife is placed in the basement and most of the background can be assumed to come from naturally occurring radioactive isotopes in the floor and the surrounding walls, made of concrete. Radioactivity in building materials, such as concrete, commonly comes from isotopes of potassium, thorium, uranium and radium. [3]

The Gamma Knife room at Sophiahemmet has relatively thin walls that have been radiation shielded with steel and the room is placed on the ground level.

The background contribution dose was estimated by measuring the activity in a similar room nearby (without a Gamma Knife machine). Here the assumption was made that the rooms were approximately the same size and the walls of the same thickness and material.

To find any background variations in the room, depending on the distance from the walls, measurements were performed with increasing distance from the wall. No larger variations were found so the measurement was therefore concentrated at a point in the middle of the room, where data was collected for a longer time (~45min). This was necessary to collect good statistics, due to the lower rate of the background radiation.

The contribution from background radiation was calculated to ~0.16 µGy/h at Radiumhemmet and ~ 0.2 µGy/h at Sophiahemmet. (See section 4.2)

4.1.4 Detector efficiency

Photons with high energies may escape the detector before all of their energy is deposited. Some of the photons may travel through the detector without undergoing any interactions at all, and leave with full energy. Thus the absorbed dose spectrum, from monoenergetic photons, will show energies spread out over a region, from zero up to full photon energy. As a consequence the measured spectrum will not coincide with the real spectrum of the photons hitting the detector.

Since the measured spectrum only correspond to the absorbed energy in the detector, it is necessary to afterwards compensate for the efficiency. This can be made by unfolding the spectrum.

We separate between the absolute efficiency, $E_{abs}$ and the intrinsic efficiency, $E_{int}$ of the detector. The absolute efficiency is defined as the ratio between the numbers of photons that are registered and the numbers of photons that are emitted by the source. The intrinsic efficiency is defined by the amount of photons hitting the detector that are actually registered and mainly depends on the scintillating material. [3]

The total detector efficiency depends primarily on the distance to the source, the properties of the detector material, such as density and optical transparency, but also on the thickness and diameter of the crystal, which determines the ratio between the different interaction processes (photoelectric effect, Compton scattering, pair production). The efficiency also varies with the photon energy, leading to the definition of photopeak efficiency, $E_{peak}$ the efficiency for producing full-energy peak pulses only.
One way to unfold the measured spectrum is to use measured or simulated spectra from single photo energies.

In this work the results from the Monte Carlo simulations are presented as energy spectra from the detector model. Therefore the simulated spectra can be directly compared with the measured ones without knowing the detector efficiency.

**Figure 4.4** Simulated detector efficiency, where the dark curve corresponds to the energy spectrum from photons hitting the detector while the light one shows the simulated energy spectrum of the scintillator.

Figure 4.4 shows the simulated effect of the detector efficiency using the Monte Carlo method described in chapter 5. For comparison the spectra are blown up and shown together with the corresponding measured spectrum in Figure 4.5.

**Figure 4.5** The simulated spectra from Figure 4.4 together with the corresponding measured spectrum. NB The simulated spectra are blown up 14 times, due to the poor statistics.
4.2 Analysis of spectra and dose calculation

The collected data and the spectra were plotted and analyzed using the MCA-software program and a Matlab code (see Appendix 8.1).

Figure 4.6 shows one of the measured spectra at Radiumhemmet, in front of the Gamma Knife 1 meter from the shielding doors and 0.5 meter off center. The full energy photons from $^{60}$Co (1.17 and 1.33 MeV) can be distinguished in the spectrum as two smaller peaks. The events registered with higher energies than 1.33 MeV are mainly so-called “sum peaks”, caused by coincident detection of two photons, and background events.

Photons that only deposit part of their energy (due to Compton scattering in the LGK and in the detector) will appear in the lower energy region of the spectrum. As can be seen in the figure the major part of the dose is delivered by low energy photons.

The absorbed dose in the detector was calculated according to the definition (see section 2.3) with following equation;

$$D = \frac{E}{m_{\text{detector}}} = \frac{E}{(V\rho)} \quad [\text{J/kg}]$$  (eq. 4.1)

where $E$ is the total energy absorbed in the detector (integrated spectrum) and $m$ is the detector mass, which can be expressed as the volume ($51\text{mm}*51\text{mm}$-cylinder) times the density ($3.67 \text{ g/cm}^3$ for NaI). Since the volume and density of the detector is known the mass can easily be calculated to 0.382 kg.
The total absorbed energy in the detector is obtained by integrating the spectrum. Using the equation above, the absorbed dose was calculated using Matlab for the different points.

4.3 Results

4.3.1 Measured doses at Karolinska Hospital

![Diagram showing absorbed dose in detector at Karolinska Hospital, 98cm above the floor given in µGy/h.](image)

**Figure 4.7** Absorbed dose in detector at Karolinska Hospital, 98cm above the floor given in µGy/h.
Figure 4.8 Absorbed dose in detector at Karolinska Hospital, 4 cm above the floor given in µGy/h.

Figure 4.9 Absorbed dose in detector at Karolinska Hospital, 192 cm above the floor given in µGy/h.
4.3.2 Measured doses at Sophiahemmet

**Figure 4.10** Absorbed dose in detector at Sophiahemmet, 98cm above floor given in μGy/h.

**Figure 4.11** Absorbed dose in detector at Sophiahemmet, 4 cm above floor given in μGy/h.
The results show that the absorbed doses at Radiumhemmet are several times higher than the ones measured at Sophiahemmet. The enhanced radiation level at Radiumhemmet is mainly due to the thinner shielding and a higher leakage between the doors.

We can also see that in some points, close to the walls at Sophiahemmet, the doses are very near or less than the measured background. This could be due to the fact that the walls shield from outside radiation.

4.4 Measurement uncertainties

One of the most probable sources of error in the results is placing the detector in the right position. The accuracy of the detector position is estimated to be in the order of centimeters. This may have considerable effects on the results, especially close to the front of the machine where the acquisition rate varies more.

According to a Monte Carlo simulation the count rate decreased almost 42% in the detector position A (see figure 5.4) by moving the detector 10cm in both x and y directions.
5. Monte Carlo simulation

5.1 The Monte Carlo technique

In contrast to conventional numerical solutions, where you need a mathematical model (normally described by differential equations) to solve a physical system, the Monte Carlo simulation technique is based on random numbers and probability statistics.

The method is very useful in calculations of radiation transport where the particle interactions can be considered as statistical processes. The physical system is described by probability density functions and the simulation is proceeded by letting the computer sample random numbers from these functions, e.g. with a random number generator. A particle track is characterized by a series of states, each state defining the particle position, energy and direction after a collision. The length between a collision, the change of direction and energy loss will be randomly sampled from the probability density functions. [6]

The random selection process is repeated many times to create multiple scenarios, so called particle histories. The statistical uncertainty of a simulation is proportional to the square root of the number of generated scenarios. Therefore the number of simulated scenarios needed to achieve a given uncertainty can be estimated by predicting the uncertainty in the average result.

To decrease the statistical uncertainties in the simulation different variance reduction methods are used, such as splitting techniques and making use of symmetries that will lower the computer time.

The Monte Carlo simulation technique is used in a wide range of fields; for example in risk analysis, dosimetry and radiotherapy, design of radiation detectors, quantum chromodynamics and stellar evolution. It is specially suited for complex and difficult problems where it is hard to apply analytical solutions. [5]

5.2 Simulation of dose distribution

The simulation of the dose distribution in this work has been performed using the Monte Carlo simulation system Pegasos. It is an algorithm and computer code for simulation of photon, electron and positron transport and based on Penelope [6].

The geometry model used in the simulation is a CAD generated model of the Leksell Gamma Knife Model C. The simulation program and model are used for particle transport and estimates of doses when using the LGK. In this work the photons will be followed from the cobalt sources, through the shielding material and out of the machine.

Before starting the simulation you need to set up a simulation input file where you specify the source and number of simulated particles, the initial state (such as position, angle and direction) and the absorption energy for photons, electrons and positrons. Setting the cutoff energy high shuts off electron transport, which will save computing time.
Apart from the geometry and simulation files you also need a material file, containing data of the different materials used.

The simulation is preceded by generating photons from Cobalt sources and letting a spherical phase space with a specified radius register the photons going through it. The result file contains information about energy, position and direction of the photons registered in the phase space.

The distance the particles have to travel through the shielding ball is long compared to the mean free path for photons in steel. Therefore only a small number of the simulated particles will reach outside the machine and the simulation could for this reason not be made in one step. Instead a number of simulations were performed, each with a larger spherical phase space, generating particles originating from the previous result file. In every simulation the number of generated particles was set so that the number of outgoing particles would be approximately the same (which could be estimated by running a shorter test simulation). See table 5.1.

A sphere of radius 45 cm was used in the first step, which result in a 5 cm distance for the particles to travel from the sources (which are located 40 cm from focus). In the following simulations 10 cm was continuously added to the radius, until the photons had reached the surface of the shielding ball.

![Figure 5.1](image-url)  
**Figure 5.1** The first three phase spaces in the simulation. The second simulation is started from the first phase space, with the photons registered in the first sphere as sources. Compare to figure 5.3, showing the particles registered in the last phase space sphere.
Figure 5.2 The geometry model of the Gamma Knife inside the room at Sophiahemmet.

For the simulations to be as true as possible, the radiation hitting and then being back-scattered from the walls should also be included in the simulations. The LGK model was therefore placed in a CAD-model of the room at Sophiahemmet (see figure 5.2) where also a model of the NaI-detector was placed in the same positions as the real detector. The simulation was then continued, letting the particles from the last phase space interact with the surrounding walls and the detector.

Since the photon interaction in the detector also was simulated, the final result shows the spectrum of the absorbed photons in the detector. These can therefore directly be compared to the measured spectra.

It should be noted that the simulation results don’t include any background radiation.
Figure 5.3 The shielding ball with the doors of the Gamma Knife. The dark dots in the picture shows photons registered in the last spherical phase space together with the particles registered in a detector positioned in front of the doors.
5.3 Results

The following detector positions were used in the simulation;

![Detector positions used in the simulation.](image)

Table 5.1 shows the data of the different steps in the simulation. The total number of primary particles, \( N \), is the number of primary particles generated from the sources multiplied with the gain factor from each step in the simulation.

<table>
<thead>
<tr>
<th>Simulation</th>
<th>Number of generated particles</th>
<th>Number of particles registered in the phase space</th>
<th>Gain factor in each simulation</th>
<th>Total number of primary particles</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Step 1</strong>- from sources to phase space with radius, ( R=45\text{cm} )</td>
<td>3.50E+09</td>
<td>6.29E+08</td>
<td>1</td>
<td>3.50E+09</td>
</tr>
<tr>
<td><strong>Step 2</strong>- from phase space ( R=45\text{cm} ) to ( R=55\text{cm} )</td>
<td>1.50E+10</td>
<td>4.74E+08</td>
<td>23.85</td>
<td>8.35E+10</td>
</tr>
<tr>
<td><strong>Step 3</strong>- from phase space ( R=55\text{cm} ) to ( R=65\text{cm} )</td>
<td>1.30E+10</td>
<td>4.75E+08</td>
<td>27.43</td>
<td>2.29E+12</td>
</tr>
<tr>
<td><strong>Step 4</strong>- from phase space ( R=65\text{cm} ) to ( R=75\text{cm} )</td>
<td>3.00E+09</td>
<td>2.99E+08</td>
<td>6.32</td>
<td>1.45E+13</td>
</tr>
<tr>
<td><strong>Step 5</strong>- from phase space ( R=75\text{cm} ) to ( R=86\text{cm} )</td>
<td>4.50E+09</td>
<td>2.37E+08</td>
<td>15.05</td>
<td>2.18E+14</td>
</tr>
<tr>
<td><strong>Step 6</strong>- from phase space ( R=86\text{cm} ) to Sophiahemmet room</td>
<td>2.50E+08</td>
<td>See table 5.2 the for different detector positions</td>
<td>1.05</td>
<td>( N=2.30E+14 )</td>
</tr>
</tbody>
</table>

*Table 5.1 Number of simulated particles in from each phase space and the corresponding number of total primary particles together with the activity and calculated gain factor, \( K \).*
To relate the registered events in the simulation to a rate (event/s) the activity (decays/s) of the sources at the day of measurement must be considered.

The LGK at Sophiahemmet was loaded in March 2002 with the total source activity 229 TBq and the time between that and the measurement is approximately 3.9 years, leading to the activity \( A = 137 \) TBq.

The conversion factor, \( K \), is calculated by dividing the present activity of the sources, \( A \), by the total number of generated primary photons, \( N \), times the number of photons from each radioactive decay;

\[
K = \frac{2A}{N} \quad [\text{particles/s}/\text{particles}] = 1.1909 \quad [/\text{s}]
\]

<table>
<thead>
<tr>
<th>Detector position in simulation</th>
<th>Number of simulated events registered</th>
<th>Simulated event rate (registered events ( \times ) ( K )) (/s)</th>
<th>Measured event rate (/s)</th>
<th>Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>8037</td>
<td>9614</td>
<td>4072</td>
<td>136.1 %</td>
</tr>
<tr>
<td>B</td>
<td>11413</td>
<td>13591</td>
<td>13272</td>
<td>2.4 %</td>
</tr>
<tr>
<td>C</td>
<td>1384</td>
<td>1648</td>
<td>1830</td>
<td>-9.9 %</td>
</tr>
<tr>
<td>D</td>
<td>5353</td>
<td>6375</td>
<td>6741</td>
<td>-5.4 %</td>
</tr>
<tr>
<td>E</td>
<td>1902</td>
<td>2265</td>
<td>2538</td>
<td>-10.8 %</td>
</tr>
<tr>
<td>H</td>
<td>2761</td>
<td>3288</td>
<td>3427</td>
<td>-4.1 %</td>
</tr>
<tr>
<td>K</td>
<td>2485</td>
<td>2959</td>
<td>1357</td>
<td>118.1 %</td>
</tr>
<tr>
<td>L</td>
<td>6899</td>
<td>8216</td>
<td>3581</td>
<td>129.4 %</td>
</tr>
<tr>
<td>A(near floor)</td>
<td>1077</td>
<td>1283</td>
<td>1092</td>
<td>17.5 %</td>
</tr>
<tr>
<td>A(near ceiling)</td>
<td>1177</td>
<td>1402</td>
<td>1172</td>
<td>19.6 %</td>
</tr>
</tbody>
</table>

**Table 5.2** Resulting event rates from the simulation in different detector position compared to the measured event rates at Sophiahemmet.

Table 5.2 shows the simulated number of events registered by the detector in the different positions, A-K, followed by the corresponding event rate (event/s). The results have been compared to the measured event rates in the real detector and the difference (%) is given in the column to the right.
Table 5.3 shows the simulated total energy of the photons absorbed by the detector. The corresponding absorbed energy per second is then obtained by multiplying with the gain factor, K. The absorbed doses in the detector are calculated using equation 4.1 and compared to the absorbed doses calculated from the measured spectra.

From the results we can conclude that the simulated count rates on the backside of the LGK are in agreement with those measured, within 15 %, while those simulated on the front side differ by several orders of magnitude from the measured. The differences are somewhat larger for the absorbed doses, but follow the same pattern; good agreement on the backside while large disagreement in front of the shielding doors.

In the back of the LGK, where the geometry is more straightforward, the photons only travel through steel and we obtain good agreement with the measured values. The photons coming out on the front side on the other hand have to travel through air before reaching the shielding doors. Since the mean free path for photons in air is longer than the distance between the phase spaces this will lead to photons going from one phase space to another without undergoing any interactions. Thus the same particles will be registered in the next phase space, unchanged in energy and direction.

The number of generated particles in every new simulation is several times higher than the number of registered particles in the previous phase space (see table 5.1) and the same particles are therefore generated many times.

These duplicated photons appear as “unwanted” peaks in the spectrum (see figure 5.5). The spectra and doses simulated in front of the Gamma Knife can therefore not be considered truthful. Figure 5.5 shows the energy spectrum from the detector in position A while figure 5.6 shows the energy spectrum from the detector in position D.
Figure 5.5 Plotted spectrum of the photon energies absorbed in the detector, in position A.

Figure 5.6 Plotted spectrum of the photon energies absorbed in the detector, in position D.

Since the number of particles shouldn’t be affected by the technique of using several phase spaces the discrepancies in the event rates is unexpected.
5.4 Simulation uncertainties

There are some uncertainties in this simulation that might have a notable effect on the results.

First the statistical uncertainty of simulation, which is proportional to the square root of the number of events. It will result in an error of a few percent, depending on the detector position.

The uncertainty of the photo-absorption coefficients in the material file lies around 2% (taken from the Penelope library from the University of Barcelona). The resulting error grows with number of interactions.

The uncertainty of the material densities is a few percent. A 2% higher density in the steel of the shielding doors can lead to 30% less radiation leakage (event rates).

The geometry file doesn’t include any other objects in the room other than the LGK.

Also physical approximations in the program, the exclusion of the electron transport and any possible differences between the geometry model and the real machine contribute to a higher uncertainty of the results.
6. Summary and conclusions

In this Thesis the energy spectrum in different positions around the Leksell Gamma Knife has been measured and the corresponding absorbed doses have been calculated. It is observed how the differences between the two models B and C affect the photon transport in the machine and thereby also the dose distribution in the treatment room. The absorbed doses measured with the detector are several times higher in the treatment room as compared with the older model.

The energy spectra from the scintillating detector show that the main part of the doses are delivered by the lower energy photons that are reduced in energy due to Compton scattering in the shielding ball.

A Monte Carlo simulation method for simulating the photon transport from the sources out of the machine has been tested and evaluated. It’s been done by calculating and comparing the simulated absorbed doses as well as the count rates to the experimental data.

The result shows that the Monte Carlo simulation technique is applicable on photon transport in the LGK while the particles travel through the shielding ball on the backside of the machine. The comparison shows agreements in spectra shapes, absorbed doses and count rates, within the uncertainty range.

In contrast, the spectra in front of the LGK do not match well with the ones measured. This suggests that the simulation technique wouldn’t be suited for photon transport through air.

The simulation method used would provide a fast and easy approach of calculating the dose distribution and the needed shielding (material and thickness) for each installation and might in the future be used e.g. in a site-planning program for the LGK. As stated in the introduction part, custom made shielding design for each LGK would most likely lower the installation cost for the buyer.

However, this simulation technique will not be applicable on a site-planning program unless the disagreement and duplication problem on the front side of the LGK can be solved. Theoretically it would be possible to perform the simulation in one step, using only one spherical phase space surrounding the shielding ball. But this would require so many generated particles that the simulation would take years to finish with acceptable statistics.
7. References

Literature

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8. Appendix

8.1 Matlab code for calculating the absorbed dose in the detector

clear

%choose file
[fname,pname] = uigetfile("*.mcd",'Select The MCD Spectrum File');

filename=strcat(pname,fname);
fid = fopen(filename,'r','l');

dummy=fread(fid,200,'int8');
data=fread(fid,8192,'int32');
dummy=fread(fid,94,'int8');

fclose(fid);
X=data(35:2048);
Y=(35:2048);

plot(Z,X); 
axis([0 2048 0 1.2*max(data(20:8180))]);
title(filename);
xlabel('keV');
ylabel('Counts');

%Calibr.eq-30sept
%Z=-19.894539+1.019792*Y);
%Calibr.eq-3okt(bakgrund)
%Z=-9.325853+1.019042*Y);
%Calibr.eq-7okt (1-30)
%Z=-8.761912+1.009506*Y);
%Calibr.eq-7okt (31-84)
%Z=-8.695318+1.015731*Y);
%Calibr.eq-14okt
%Z=-8.598994+1.020230*Y);
%Calibr. Sophia
Z=-8.411159+1.022667*Y);

plot(Z,X); 
axis([0 2048 0 1.2*max(data(20:8180))]);
title(filename);
xlabel('keV');
ylabel('Counts');
%Calib. Sophia
\[ v = -8.41159 + 1.022667 \times (35:2048); \]

%AREA (keV/100s)
\[ k = \text{data}(35:2048); \]
\[ s = v \times k \]

%JOULE
\[ J = s \times (1.6022 \times 10^{-16}); \]

%Absorbed Dose in Detector (µJ/kg/h)
\[ r = 5.1, \] %cm
\[ V = \frac{r^3}{4\pi}, \] % cm³
\[ p = 3.67, \] % g/cm³
\[ m = V \times p / 1000, \] % kg
\[ D = \frac{J}{m} \times 36 / 10^{-6} \] % (J/kg)µ/h