Measurements and Monte Carlo Simulations of Neutron Doses from Radiation Therapy with Photons, Protons and Carbon Ions

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Abstract

During radiation therapy for cancer, patients receive undesired dose from neutrons produced in collimators and in the patient. Dose from neutrons is associated with a potentially increased risk of secondary malignancies after radiation therapy.

The overall objective of this thesis has been to investigate the magnitude and distribution of neutron dose from radiation therapy with photons, protons and carbon ions. The pursuit of this ambition has also required an extensive study of the applied neutron detectors’ properties.

Measurements in proton and carbon ion beams were performed with a novel neutron detector based on radiation effects in Static Random Access Memory (SRAM) chips. For measurements of neutron dose in photon therapy, bubble detectors and thermoluminescence detectors (TLDs) were applied.

Monte Carlo simulations with the FLUKA Monte Carlo simulation package were conducted for comparison with the experimental data. Characterization experiments with SRAM detectors were conducted at PTB (Physikalische Technische Bundesanstalt) in Braunschweig, at TSL (The Svedberg Laboratory) in Uppsala, at CERF (the Cern Eu Reference Field) outside Geneva, at the OCL (Oslo Cyclotron Laboratory) in Oslo, at Haukeland University Hospital in Bergen, and at the Jeep II research reactor at the Institute for Energy Technology, Kjeller.

At TSL neutron dose generated from 178 MeV protons was measured with SRAM detectors placed in a water phantom. The proton beam was applied without the use of any collimators or beam shaping components. At Bragg peak depth, the experimental results from the proton beam measurements indicate neutron doses from 0.1 pSv per proton at 5.2 cm lateral distance from the beam axis to 0.02 pSv per proton at 13.7 cm lateral distance. For a 3 x 3 x 3 cm³ target volume these values may be equivalent to 0.7 mSv/Gy and 0.1 mSv/Gy, respectively. At the depth of the Bragg peak, results
from Monte Carlo simulations indicate neutron doses from 2 mSv/Gy at the beam axis, to approximately 0.1 mSv/Gy at 13.7 cm lateral distance from the beam axis. Further, the measurements suggest that the neutron dose changes little with depth in the range from two centimeter prior to the Bragg peak to three cm deeper than the Bragg peak. The Monte Carlo simulations support these findings. In addition, the Monte Carlo simulations suggest that the neutron dose increases with depth in a build-up region of about five centimeters. Beyond this build-up region, only moderate changes in the neutron fluence are observed before approaching Bragg peak depth where the neutron fluence starts to decrease with depth.

The Monte Carlo simulations indicate that the neutron dose from 341 MeV/u carbon ions (same range as 178 MeV protons) are on the order of 10 mSv/Gy at the Bragg peak and further that it is reduced to approximately 0.1 mSv/Gy at 13.7 cm lateral distance from the beam axis.

The neutron dose from photon therapy was measured in a water phantom irradiated with photons from a Varian 23 iX linear accelerator in the 15 MV energy mode. The measurements from photon beams indicate that the neutron doses outside the treatment field range from 5 mSv/Gy at the surface of the water phantom, to 0.2 mSv/Gy at 10 cm depth in water. The Monte Carlo simulations suggest a range for the neutron doses outside the treatment field between 0.8 mSv/Gy at the surface of the water phantom and 0.1 mSv/Gy at 10 cm depth in water. Both measurements and Monte Carlo simulations show that the neutron dose in photon therapy is strongly dependent on the depth in the patient, and is only to a small extent dependent on the lateral distance from the treatment field.

In summary, the measurements and Monte Carlo simulations indicate that close to the beam axis, the neutron dose from protons and ions may be higher than the neutron dose from photons. With increasing lateral distance from the beam axis the neutron dose from photons may be expected to at some point surpass the neutron dose from protons and carbon ions. At which distance this occurs will depend on the depth.
considered. The results in this work indicate that at 8.6 cm depth, the neutron dose from photons will surpass the neutron dose from protons and carbon ions at a lateral distance of approximately five to ten centimeters from the beam axis.

The results from this work may be used as a platform from which to further investigate and quantify the neutron field around particle therapy beams and medical linear accelerators.
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1. Out-of-field dose in radiation therapy

Since Wilhelm Röntgen's discovery of X-rays in 1895, radiation has been used to cure malignancies in the body. Radiation therapy has become one of the most valuable tools in the fight against cancer, with photons as the most common type of radiation. Photon therapy has developed and been refined continuously from the use of radioactive sources to today's high-energy photon beams produced by linear accelerators.

Photon therapy is still the cornerstone of radiation therapy, but a promising alternative has slowly emerged: The use of high energetic ions for medical purposes. The use of nuclear particles in a clinical context is often referred to as particle therapy. This concept was first proposed by Robert R. Wilson in 1946 (Wilson, 1946). The very first patient was treated with protons at Berkeley in 1954. During the last two decades the development in particle therapy has rapidly progressed, and a transfer of knowledge from nuclear physics institutions to dedicated radiation therapy facilities has taken place. Moreover, with the introduction of image guided radiation therapy (IGRT) and modern beam delivery techniques it is now possible to better utilize the potential of particle therapy. As the particle therapy technology has matured, new treatment centers have been established at an increasing rate in recent years. Figure 1.1 shows the current established and planned particle therapy facilities (PTCOG, 2013).

In radiation therapy, dose will always be deposited both in the tumor and unavoidably in healthy tissue. During planning and optimization of a radiation treatment, delivering the prescribed dose to the tumor and reducing dose to organs and tissue close to the target volume will normally be the priority issues. The low-dose regions far from the treatment volume have usually received little attention. A part of the reason for this is that there have been no treatment alternatives available for reducing the dose in these regions.
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Figure 1.1: An overview of existing and planned particle therapy centers world-wide. Some of the carbon ion facilities do also have access to protons. Data from (PTCOG, 2013).

The present establishment of particle therapy in the clinics worldwide now enables a general opportunity to notably reduce the integral dose to the patients by the use of the more favorable dose distributions associated with particle therapy.

With the increasing survival rates of modern radiation therapy, and the rising number of cancer survivors with prospects of an improved quality of life and a long post treatment life, the long term effects from radiation may become more evident. In addition, modern conformal treatment techniques such as Intensity Modulated Radiation Therapy (IMRT) and passive proton therapy may elevate radiation exposures in areas distant from the target volume (Xu et al., 2008). In this context, a need to thoroughly investigate the radiation doses to healthy tissue in the low-dose regions has emerged.

Several review studies emphasize the relevance of the low dose regions for the risk of late effects, in particular secondary cancer (Newhauser and Durante, 2011, Miralbell et al., 2002, Xu et al., 2008). Still, the long term consequences of low dose exposures are not well known, and many questions remains (BEIR, 2006). Due to the
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fundamental physical differences between radiation treatments with photons and ions, the out-of-field dose distribution may be expected to differ significantly.

Particle therapy with active beam delivery techniques has the advantage that the only significant source of secondary radiation is interactions between the beam particles and the patient itself. In photon therapy, neutrons are produced in collimators and other components of the treatment head of the accelerator. Neutron production is associated with the modulation of the particle beam upstream and relatively close to the patient. This is also an issue in particle therapy when passive beam scattering techniques are applied. However, in passive particle therapy the neutrons produced internally in the patient also play an important role (Clasie et al., 2010, Jiang et al., 2005).

The figures presented in this thesis is from work performed by the author unless otherwise is specified. All the results from Monte Carlo Simulations are based upon work by the author. Figure 1.2 shows results from Monte Carlo simulations of

![Relative neutron dose](image)

Figure 1.2 - Monte Carlo simulations: Relative distribution of neutron dose on the surface and inside water phantoms irradiated with a 3 x 3 cm² photon field in the 15 MV energy mode (right) and a carbon ion irradiation of a 3 x 3 x 3 cm³ target volume (left). The radiation fields enter the center of the cube from above. In photon therapy neutrons from the collimators in the treatment head are distributed over large areas and are moderated with depth. With the carbon ion beam neutrons are primarily created along the primary beam or field. In order to display the neutron dose inside the water phantom, one half of the phantom has been made transparent.
neutron dose from therapeutic beams of carbon ions (left) and photons (right). This result illustrates the wide distribution of neutrons from collimators in photon therapy while, for the carbon beam, which is applied without collimation, the neutrons are produced in interactions inside the patient along the beam axis.

Quantification of the dose in the low-dose regions will give a starting point for the assessment when considering the benefits and disadvantages of different treatment modalities with respect to late effects far from the tumor site, such as secondary malignancies.

1.1 Interaction mechanisms between radiation and matter

1.1.1 Photons

The photon is a massless, electrically neutral, particle that interacts through electromagnetic interaction mechanisms. The photon is defined as the gauge particle for electromagnetic interactions in the standard model for particles and matter. This means that it is the carrier of the electromagnetic forces in the universe.

Photons may undergo a multitude of different interactions when passing through matter. The main interactions are however: the photoelectric effect, Compton scattering and pair production (Leo, 1994). Other less dominant interactions are Rayleigh scattering and photonuclear reactions.

The photoelectric effect

A photon passing through matter may be absorbed by an atomic electron, and the electron is subsequently ejected from the atom (Khan, 1994). This phenomenon is known as the photoelectric effect, and the process is illustrated in Figure 1.3. The energy from the photon is transferred to the electron, consequently, the kinetic energy of the outgoing electron will be:
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\[ E_k = E_\gamma - E_b, \]

where \( E_\gamma \) is the photon energy and \( E_b \) is the binding energy of the electron. After such a process, there will be an electron missing from one of the electron shells. An electron from one of the outer shells will undergo a transition to fill this vacancy, and at the same time it will emit characteristic x-rays (Khan, 1994). There is also the possibility of emission of Auger electrons, which are monoenergetic electrons produced by the absorption of characteristic x-rays internally by the atom. For photons with energies in the MeV range the cross section for the photoelectric effect increases with \( Z^4 \) or \( Z^5 \), \( Z \) being the atomic number of the host material (Leo, 1994). This indicates that high-Z materials are efficient as shielding materials for photons.

![Diagram of the photoelectric effect](image)

*Figure 1.3: The figure illustrates the photoelectric effect. An incoming photon is absorbed by an electron which in turn escapes from the atom. Electrons occupying higher shell states will fall to a lower shell to fill the vacancy. Characteristic x-rays are emitted in this latter process.*

**Compton scattering**

Compton scattering is the scattering of photons on free electrons (Leo, 1994). In matter, the electrons are bound, but if the photon energy is high compared to the binding energy of the electrons, the electrons can be considered to be virtually free. In Compton scattering the incoming photon is scattered on an atomic electron and a part of the photon's energy is transferred to the electron leaving the atom ionized (Figure 1.4).

Because the electrons are considered to be essentially free, the electron density is the most important factor for attenuation of a photon beam by Compton scattering.
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![Figure 1.4: An illustration of the Compton scattering process. A photon is scattered on an atomic electron and a part of the photon's energy is transferred to the electron leaving the atom ionized.](image)

Photons with energies in the MeV range will deposit a larger fraction of their energy through Compton scattering compared to keV photons, and as Figure 1.6 shows, photon interactions are dominated by Compton scattering in the MeV range.

Related to Compton scattering is the classical process of Rayleigh scattering. Rayleigh scattering is scattering of photons by atoms as a whole. This is also referred to as coherent scattering because all the electrons in the atom participate in a coherent manner. In Rayleigh scattering the atoms are neither excited nor ionized and only the direction of the photon is changed.

**Pair production**
The mechanism of pair production involves a photon transforming into an electron-positron pair as seen in Figure 1.5. All the photon energy is transferred to the created particles. The electron has a rest mass of 0.511 MeV. Consequently, the threshold photon energy for pair production is 1.022 MeV. The photon energy excess above this threshold is distributed between the two particles, and the most probable case is for each particle to acquire half of the excess energy. In order to conserve momentum, pair production can only occur if there is another particle in the vicinity, usually a nucleus (Leo, 1994). Contrary to the case for the Compton effect, the probability for pair production increases rapidly with the atomic number of the material.
Photonuclear reactions

A photon can interact with a nucleus resulting in the emission of one or more of the nucleons. The photon energy must exceed the binding energy of ejected nucleons. In general the thresholds for (γ,p) reactions are higher than for (γ,n) reactions due to the Coulomb barrier that a proton must overcome to escape the nucleus (Turner, 2009). Neutrons are produced from photons through two different interactions; evaporation and the direct knockout effect (Zanini et al., 2004). When a nucleus as a whole is excited by a high energy photon, a neutron can be emitted (evaporated) in the de-excitation process. The angular distribution of the emitted neutrons is isotropic. In the case of the direct effect, a photon interacts with the neutron in a nucleus and the
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Photon energy is absorbed by the neutron resulting in the neutron being knocked out of the nucleus. The angular distribution of the emitted neutrons is forward peaked.

Above the threshold energy, the cross section for neutron production increases with photon energy up to a maximum value and then decreases again with higher photon energies (NCRP, 1984). This peak in the cross section is known as the giant dipole resonance (GDR). In Figure 1.7 the Giant Dipole Resonance for tungsten ($^{186}$W) is shown. The cross section is highest for energies between 10 and 20 MeV. However, as the figure also shows, the photonuclear cross section is small compared to the total photon interaction cross section.

![Figure 1.7 - Measurements: The giant dipole resonance (GDR) and the total photon interaction cross section in Tungsten ($^{186}$W). The GDR can be seen reaching its maximum at 15 MeV. Data from (Chadwick et al., 2011) and (Kishida et al., 2004).](image)

**Photon beam attenuation**

As a photon beam passes through matter its intensity is continuously reduced. The change in intensity can be expressed as:

$$dl = -\mu dx.$$  \hspace{1cm} (1.2)

Where $\mu$, the total absorption coefficient, is the constant of proportionality and $I$ is the beam intensity. Solving equation (1.2) yields:
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\[ I(x) = I_0 e^{-\mu x}, \]  

(1.3)

where \( I(x) \) is the intensity after passing through a thickness \( x \) of the material and \( I_0 \) is the initial photon intensity.

When a photon beam traverses matter, the dose deposition will increase with increasing depth in a build-up region up to the maximum dose located a few cm inside the tissue. The region in front of the dose maximum position is known as the build-up region because an increasing number of secondary electrons are set in motion by the photons. Due to the exponential decrease in the photon beam intensity with depth and also due to the approaching of the equilibrium condition between the number of released and produced electrons traversing into a volume and the number of electrons exiting that volume, the dose deposition eventually begins to decrease. The location of the dose maximum will depend on the beam energy as seen in Figure 1.8.

![Depth dose curves for photons](image)

Figure 1.8 - Measurements: Depth dose curves measured for the beam qualities 15 MV and 6 MV produced by a medical linear accelerator at Haukeland University Hospital.

1.1.2 Heavy charged particles

**Energy loss and stopping power**

Charged particles traversing matter lose energy primarily through inelastic collisions with atomic electrons causing excitation and ionization of the atoms. A heavy
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charged particle, like the proton or a carbon ion, transfers only a small fraction of its energy in a single collision with an atomic electron, and is only to a small degree deflected from its initial path. Thus, a heavy charged particle can propagate along an almost straight path through tissue leaving ionized and excited atoms along the track. The particle energy, and thus the velocity, is reduced by a small amount in each collision.

Energy loss of charged particles by ionization and excitation of atoms in matter is described by Equation (1.4), the Bethe-Bloch formula for heavy charged particles (Leo, 1994):

$$\frac{dE}{dx} = 2\pi N_r r_e^2 m_e c^2 \rho \frac{Z Z^2}{A \beta^2} \left[ \ln \left( \frac{2m_e \gamma^2 v^2 W_{max}}{I^2} \right) - 2\beta^2 - \delta - 2 \frac{C}{Z} \right]$$  (1.4)

The Bethe-Bloch formula indicates that the range of particles is proportional to $A/Z^2$, where $A$ is the atomic weight and $Z$ is the charge of the particle. Furthermore, as the velocity decreases, the energy loss per length will increase resulting in maximum energy deposition at the end of the particles range. The energy loss per length, dE/dx, is also referred to as the stopping power. In Figure 1.9, both the stopping power and the dose (energy) deposition of protons in water is shown. The plot illustrates the energy dependence of the stopping power, and thus dose deposition, as described by the Bethe-Bloch formula. The terms in the equation are explained further in Table 1.1.

At low energies (of the order of 10 MeV/u or less) the Bethe-Bloch equation is no longer valid because the ions start to pick up electrons from the traversed material. As the charge of the ions is reduced, the energy loss will also change accordingly. The ions will finally come to rest as neutral atoms.

An important quantity related to the stopping power is the Linear Energy Transfer (LET). The LET is the energy deposited locally per unit path length and differs therefore from the stopping power because the LET does not include energy loss.
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Table 1.1: Terms in the Bethe Bloch formula.

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>$r_e$</td>
<td>Classical electron radius</td>
</tr>
<tr>
<td>$m_e$</td>
<td>Electron mass</td>
</tr>
<tr>
<td>$N_a$</td>
<td>Avogadro’s number</td>
</tr>
<tr>
<td>$z$</td>
<td>Charge of incident particle in units of e</td>
</tr>
<tr>
<td>$\beta$</td>
<td>$v/c$ of the incident particle</td>
</tr>
<tr>
<td>$\gamma$</td>
<td>$1/\sqrt{1-\beta^2}$</td>
</tr>
<tr>
<td>$I$</td>
<td>Mean excitation potential</td>
</tr>
<tr>
<td>$Z$</td>
<td>Atomic number of absorbing material</td>
</tr>
<tr>
<td>$A$</td>
<td>Atomic weight of absorbing material</td>
</tr>
<tr>
<td>$\rho$</td>
<td>Density of absorbing material</td>
</tr>
<tr>
<td>$\delta$</td>
<td>Density correction</td>
</tr>
<tr>
<td>$\gamma$</td>
<td>Shell correction</td>
</tr>
<tr>
<td>$W_{\text{max}}$</td>
<td>Maximum energy transfer in a single collision</td>
</tr>
</tbody>
</table>

Figure 1.9: Left: Total stopping power of protons in water as a function of energy. Right: A plot of the dose deposition from a 107 MeV proton beam on water as a function of depth. The plots illustrate how the energy deposition of the protons increases with decreasing energy. Stopping power data from (NIST, 2013). The depth dose was obtained through FLUKA Monte Carlo simulations.

through bremsstrahlung. The LET value gives a measure of the density of ionizations induced by the beam particles, and it is proportional to the square of the electrical charge of the beam particles, thus indicating that carbon ions will have higher LET values than protons. Further, the LET is inversely proportional to the ion energy. Consequently, the LET of the beam particles is higher in the Bragg peak compared to in the entrance channel of the beam.

**Lateral spread and inelastic collisions**

Particle and photon beams will always be broadened as they pass through matter. This is due to elastic interactions between the beam particles and nuclei in the
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material, often referred to as multiple Coulomb scattering (Ma and Lomax, 2013). The scattering is proportional to the charge of the projectile and inversely proportional to the atomic weight and velocity of the projectile. This results in a more than three times larger broadening of a proton beam compared to a carbon beam with the same range (Weber and Kraft, 2009). Figure 1.10 illustrates the lateral scattering in water and shows that proton beams are more laterally scattered than carbon ion beams. The plots show the absorbed dose in water, and are obtained through FLUKA Monte Carlo simulations. Dose levels to 0.1% of the maximum dose (100%) are shown in the figure.

![Figure 1.10 - Monte Carlo simulations: The figure illustrates the lateral spread of proton and carbon beams as a function of depth in water. In addition, the dose tail from fragmentation of carbon ions is visible. The color map indicates the absorbed dose levels from 100% dose in the Bragg peak down to doses as low as 0.1%. The blue-grey background illustrates the water phantom.](image)

Heavy charged particles may create secondary particles through inelastic interactions with matter. Ions such as carbon ions will with a certain probability be fragmented into lighter ions when passing through matter and may also fragment nuclei in the medium producing secondary charged ions and neutrons.
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Beam fragments have a forward peaked distribution due to the high velocity of the primary beam particles, and this leads to a build-up of lower Z fragments with increasing depth (Gunzert-Marx et al., 2008). Due to their lower charge, the fragments may have longer ranges than the primary ions. Consequently, the depth dose distribution of a heavy ion beam includes a small dose tail, in the distal end, beyond the range of the primary ions. This effect can be seen in Figure 1.10. Neutrons and other fragments originating from the target show an isotropic angular distribution and are emitted with much lower velocities than the fragments from the beam particles.

1.1.3 Neutrons

Due to the neutrons lack of electrical charge, their interaction with matter differs significantly from that of charged particles. Neutrons interact with nuclei in the material they traverse, and the secondary radiations from neutron interactions are therefore often heavy charged particles.

The relative probabilities for different neutron interactions are strongly dependent on the energy of the neutrons (Knoll, 2010). Neutrons with energies above a few hundred keV are often referred to as fast neutrons. Elastic scattering from nuclei is the primary mechanism of energy loss for fast neutrons in the MeV range (Turner, 2009). In elastic scattering, the total kinetic energy is conserved in the interaction, and hence, the energy lost by the neutron is equal to the kinetic energy of the recoil nucleus. When energy from neutrons is transferred to charged particles through scattering, a light nucleus will in general absorb more of the neutrons’ energy than heavier nuclei. This is the reason why hydrogenous materials such as water and paraffin (CH₂) are effective neutron moderators.

Inelastic collisions between fast neutrons with nuclei will also occur. In this case a nucleus absorbs some energy internally and is left in an excited state. For inelastic collisions to occur, the neutron must have sufficient energy to excite the nucleus.
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Therefore, only elastic scattering is observed below energies on the order of 1 MeV (Leo, 1994). However, with decreasing neutron energy the probability of another type interaction, neutron capture, increases. In this case, the neutron is captured or absorbed by a nucleus leading to a reaction such as (n,p) or (n,γ). The cross section for neutron capture is proportional to 1/ν where ν is the velocity of the neutron. Neutrons with energy on the order of 0.025 eV are said to be thermalized because they are in thermal equilibrium with the surrounding atoms (Leo, 1994). Neutrons may be captured before they are thermalized, but at thermal energies, the neutrons will diffuse through matter until they are captured by a nucleus or until a nuclear reaction takes place. Examples of important neutron capture reactions in tissue are $^1\text{H}(n,\gamma)^2\text{H}$ and $^{14}\text{N}(n,p)^{14}\text{C}$.

Neutron depth dose curves from Monte Carlo simulations are shown in Figure 1.11. For the low neutron energies, the dose decreases rapidly with depth, while for higher energies, the dose deposition remains high also for large depths. The Monte Carlo simulations show that the depth dose distribution for neutrons is strongly dependent on energy.

![Neutron depth dose curves](image)

**Figure 1.11 - Monte Carlo simulations: Neutron depth dose curves in water for four different initial energies. These curves illustrate how the neutron dose may be differently moderated with depth in photon therapy and particle therapy due to the large difference in primary beam energy, and thus secondary neutron energy.**
1.2 Radiation therapy

The goal of radiation therapy is to kill tumor cells while minimizing the damage to the rest of the body. The quality of a radiation treatment can be reflected by the parameters tumor control probability (TCP) and normal tissue complication probability (NTCP). The TCP is dependent on the dose delivered to the tumor. On the other hand, the NTCP also depends on the radiation dose, i.e. the dose to normal tissue. The ideal situation from a radiation therapy perspective would be to deliver the prescribed dose to the target volume and zero dose to the normal tissue. Unfortunately an increased dose to the tumor will increase the dose deposited in healthy tissue due to the depth dose distributions of all available radiation sources. A compromise must be made between the TCP and the NTCP in such a way that the treatment gives the patient the best achievable quality of life for the future. The possibility to achieve good local control while keeping the NTCP low is illustrated in Figure 1.12 and is often referred to as the therapeutic window (Holthusen, 1936).

![The therapeutic window](image)

*Figure 1.12: The figure shows dose response curves for tumor control probability and normal tissue complication probability. The gap between the two curves shows the potential for curing the patient while keeping side effects low and is referred to as the therapeutic window (Holthusen, 1936).*

The most common modality of radiation therapy today is external beam photon therapy, while particle therapy has emerged as promising alternative. Independent of
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the treatment technique considered, the developments in radiation therapy have as ambition to expand the therapeutic window.

1.2.1 Dosimetric principles and quantities

Dosimetry is the science of quantifying radiation in order to normalize radiation doses, to maintain calibration standards and to enable predictions of the consequences from an exposure to radiation. A vast number of different definitions are used, some based on energy deposition in a medium while other quantities aim to describe the biological effect on the human body.

**Absorbed dose**

The absorbed dose, \( D \), is defined as the energy, \( \Delta E \), which is deposited per mass unit in a small mass, \( \Delta m \):

\[
D = \frac{\Delta E}{\Delta m},
\]

(1.5)

The unit for absorbed dose is the Gray [Gy].

**Equivalent dose and effective dose**

The equivalent dose, \( H_T \), gives a normalized measure of the biological effect from irradiation of tissue or an organ. It takes into consideration which type of radiation the tissue, or an organ is exposed to, and it is calculated by multiplying the absorbed dose, averaged over the tissue or organ (\( D_R \)), by the appropriate radiation weighting factor (\( w_R \)):

\[
H_T = w_R \times D_R,
\]

(1.6)

The variation in biological effect from different particles, and therefore also the weighting factors, is related to the difference in linear energy transfer (LET). The unit for equivalent dose is the Sievert (Sv), and weighting factors from the International Commission on Radiological Protection (ICRP) are listed in Table 1.2.
Out-of-field dose in radiation therapy

Table 1.2: Radiation weighting factors from the ICRP publication 103 (ICRP, 2007).

The entity $E_N$ represents the neutron energy.

<table>
<thead>
<tr>
<th>Radiation type</th>
<th>Radiation weighting factor, $w_R$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Photons</td>
<td>1</td>
</tr>
<tr>
<td>Electrons and muons</td>
<td>1</td>
</tr>
<tr>
<td>Protons and charged pions</td>
<td>2</td>
</tr>
<tr>
<td>Alpha particles, fission fragments and heavy ions</td>
<td>20</td>
</tr>
</tbody>
</table>

Neutrons:

$E_N < 1$ MeV:  
\[ 2.5 + 18.2e^{-\frac{[\ln(E_N)]^2}{6}} \]

$1$ MeV < $E_N$ < $50$ MeV:  
\[ 5.0 + 17.0e^{-\frac{[\ln(2E_N)]^2}{6}} \]

$E_N > 50$ MeV:  
\[ 2.5 + 3.25e^{-\frac{[\ln(0.04E_N)]^2}{6}} \]

Whole body exposures to radiation are rarely uniform. The equivalent dose received may vary considerably from one organ to another. Also, the effect of radiation is found to depend on the specific organ or tissue receiving the radiation. To take this into account, dimensionless tissue weighting factors, $w_T$, are defined for various organs of the body. The effective dose, $E$, is defined as:

\[ E = \sum_T w_T \times H_T. \]  

(1.7)

The effective dose is, like the equivalent dose, measured in Sieverts.

The equivalent dose and effective dose are known as limiting dose quantities applied as occupational dose limits for personnel working with radiation. From Table 1.2 we see that the radiation weighting factor for ions is set to a high value of 20, but the strong energy dependence of the LET and thus the energy dependence of the biological effect has not been taken into account. This illustrates that these quantities are intended to provide guidance in approximating potential harmful effects rather than to give deterministic biological risk estimates.
**Operational dose quantities**

Effective- and equivalent doses are not directly measurable because they are defined in the body of each individual. The International Commission on Radiation Units and Measurement (ICRU) has therefore defined the personal and ambient dose equivalent. These are so-called operational dose quantities (ICRU, 1993) and are intended for use in practical measurements.

For all types of radiation the operational quantities for area monitoring are defined on the basis of a phantom, the ICRU sphere. It is a sphere with a diameter of 30 cm made from tissue-equivalent material (Figure 1.13). It adequately approximates the human body with regard to scattering and attenuation of the radiation fields under consideration (Wernli, 2004).

For area monitoring of penetrating radiation (including neutrons) the operational quantity is the ambient dose equivalent, $H^*(d)$, with $d = 10$ mm ($H^*(10)$) (ICRU, 1993). The ambient dose equivalent, $H^*(10)$ is defined as the dose equivalent that would be produced by the corresponding aligned and expanded radiation field, in the ICRU sphere at a depth 10 mm, on the radius vector opposing the direction of the aligned field, as seen in Figure 1.13. The unit of ambient dose equivalent is Sievert. $H^*(10)$ gives in most cases a conservative estimate of the effective dose a person would receive (Wernli, 2004).

*Figure 1.13: Illustration of the geometrical definition for measurement of the operational dose quantity, $H^*(10)$, ambient dose equivalent for strongly penetrating radiation. Modified from (Bedogni, 2006).*
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**Relative biological effectiveness (RBE)**

While the effective dose quantity aims to give a conservative estimate of the risk from radiation, and thus represent an upper limit for use e.g. in radiation protection, the quantity Relative Biological Effectiveness (RBE) was introduced to more accurately estimate the biological effect of radiation. The concept of the RBE is applied in order to achieve sufficient accuracy in ion therapy treatment planning (Schardt et al., 2010).

Mathematically the RBE value for ion therapy is defined as the ratio of dose from X-rays to ion dose producing the same biological effect (Kraft, 2000a):

\[
RBE = \frac{D_X}{D_{ion}}.
\]  

(1.8)

In dose calculations the biological effective dose (BED) can be calculated by multiplying the absorbed dose with the relevant RBE factor. The resulting BED is usually expressed in the unit Gray equivalent (GyE), and ideally, this value represents the photon dose needed to achieve the same biological effect as the particle radiation.

The RBE values depend on many different parameters including particle type and energy, dose and which type of tissue is considered. In carbon ion therapy, the treatment planning therefore relies on complex biological models such as the Local Effect Model (LEM) developed at GSI/Heidelberg (Scholz and Kraft, 1996).

Radiation can cause various forms of damage in a human cell, but damage in the cells’ DNA, which contains the complete genetic information, is the most important. Damage in the DNA may prevent the survival or reproduction of the cell. The microscopic distribution of dose determines how effectively the DNA in the cells is damaged. Densely ionizing (high LET) radiation such as carbon ions will produce more clustered DNA damage which is difficult for the cells’ DNA-repair mechanisms to handle. Figure 1.14 illustrates the difference in ionization density between carbon ions and protons and also shows how the interaction rate increases from beam
1 Out-of-field dose in radiation therapy

particles with energy 10 MeV/u to 1 MeV/u. The larger fraction of high LET particles in the Bragg peak compared to more shallow depths leads to a favorable increase in the RBE with depth when using carbon ions. The red line in the figure shows the trajectory of the primary ions while the blue lines are tracks from secondary electrons. The resulting tracks from 100 primary ions are shown in each of the four panels of Figure 1.14.

![Figure 1.14](image)

*Figure 1.14 – Monte Carlo simulations: The figure shows particle trajectories from ions (horizontal red lines) at energies 10 MeV and 1 MeV and trajectories of the electrons they liberate (blue lines) in water. The DNA as the radiation sensitive target in human cells is illustrated with a width of 2.5 nm. The resulting tracks from 100 primary ions are shown in each of the four panels.*

The biological effect of radiation has also been studied in cell experiments in order to quantify and describe the difference in cell survival after irradiation with carbon ions, protons and photons. For typical energies in the entrance channel of a carbon treatment, the cell survival, and thus the biological effect of the radiation, is found to be close to that of photons (Kraft, 2000a). In the last 2 cm of the carbon ions range, the Bragg peak area, the cell survival rate is found to be significantly lower corresponding to RBE values of about 3.
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For protons, cell experiments have shown that high biological effectiveness is achieved only for the last few micrometers of the particles range. In clinical use, the RBE for protons is typically set to a constant value of 1.1 (Paganetti et al., 2002).

Neutrons produce high-LET recoil ions as they are moderated in tissue and are therefore expected to have a high RBE. In the context of radiation induced cancer, there are many open questions regarding the RBE of neutrons. This is illustrated in an article by Brenner and Hall from 2008 where a neutron RBE of 25 was assumed for neutron carcinogenesis at low doses (Brenner and Hall, 2008). The authors emphasize the large uncertainties which, based on available data, were estimated to a factor of four in each direction. This implies possible RBE factors up to 100, and in fact, RBE values of 100 and more have been proposed for several radiation endpoints (NCRP, 1990, Brenner and Hall, 2008).

1.2.2 Photon therapy

Modern radiation therapy with photons is mainly conducted with radiation produced by a linear accelerator (linac). Electrons are accelerated to therapeutic energies, and in the treatment head of the linac, the electron beam produces photons through the bremsstrahlung process in a target of a high-Z material such as tungsten. Subsequently, a well-defined photon field is shaped by collimators. Figure 1.15 shows the major components of a typical treatment head.

For many decades radioactive cobalt was the source of the radiation in photon therapy. The energies of photons emitted from cobalt-60 are 1.17 MeV and 1.33 MeV. With the introduction of linear accelerators higher photon energies has been introduced. In the past decades the energy of the beam used in clinical practice has become as high as 25 MV, with 6 MV and 15 MV being the most frequently used beam energies. The term MV refers to the voltage applied to produce the electron beam and represents the maximum energy of the photons. The rationale for moving
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![Diagram of a linear accelerator treatment head with labels for Target, Primary collimator, Flattening filter, Ion chambers, X and Y collimators, Multi leaf collimators, and Photon field.]

**Figure 1.15:** A sketch of the major components in the treatment head of a linear accelerator. The primary collimator with the target is followed by a flattening filter, ionization chambers, collimators for the x and y direction and finally multi leaf collimators.

Towards higher beam energies is that it offers a more advantageous depth dose distribution.

Current radiation treatment (RT) techniques with medical linear accelerators include 3D-conformal RT (3D-CRT), (IMRT), Volumetric Modulated Arc RT (VMAT) and TomoTherapy. The 3D-conformal technique is based on static radiation field from several directions. IMRT includes the use of dynamic Multi Leaf Collimators (MLCs)

![Diagram illustrating VMAT showing dose distribution from three rotating fields for a treatment plan for cancer in the central nervous system. The red dots indicate the extension of the arc fields.](image)

**Figure 1.16:** The figure illustrates the principle of VMAT showing the dose distribution from three rotating fields for a treatment plan for cancer in the central nervous system. The red dots indicate the extension of the arc fields. Figure courtesy of Camilla H. Stokkevåg.
Out-of-field dose in radiation therapy

which modulates the radiation intensity of the field during treatment. With the introduction of VMAT, patients can now be irradiated also during rotation of the gantry. In Figure 1.16 a VMAT treatment plan for cancer in the central nervous system is illustrated. TomoTherapy is another variation of IMRT which includes a treatment beam rotating around the patient during irradiation.

In a medical linac, the radiation output is measured by ionization chambers placed in the photon beam in the upper part of the treatment head (see Figure 1.15). The ionization chambers measure the amount of charge released by the photon beam, and this is a measure of the photon beam's ability to ionize matter. The response from the ionization chambers is given in MU (Monitor Units), and the measured ionized charge is proportional to the deposited dose in a phantom in a reference set-up. The measured dose in this reference set-up is further applied as the link between the number of monitor units irradiated and the resulting deposited dose in a patient. For an IMRT treatment, the number of monitor units applied may be significantly higher than in conventional treatment techniques. This implies that the production of neutrons will be higher in an IMRT treatment. It has been found, independent of the treatment technique, that the neutron production is proportional to the amount of radiation output, reflected proportionally by the amount of monitor units applied (Howell et al., 2005). VMAT dose plans often require less MU compared to IMRT plans (Foroudi et al., 2012) and are therefore expected to reduce the integral neutron dose to the patient. However, studies have shown that simply scaling the neutron dose with the number of MU may not always give an accurate estimate of the neutron dose (Halg et al., 2012).

1.2.3 Particle therapy

Particle therapy is today a rapidly growing branch in the community of radiation therapy world-wide. The depth doses from protons and ions are characterized by a relatively low entrance dose and a narrow dose maximum in the Bragg peak followed by a very sharp dose fall-off beyond the Bragg peak. This favourable distribution of
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dose with depth is the main rationale for particle therapy. Through these properties, particle therapy displays a potential to substantially reduce the integral dose to the patient compared to conventional radiation therapy. Quantitatively, the integral dose from proton or ion therapy may typically be a factor of 2–3 lower than a comparable photon treatment (Lomax et al., 1999).

Figure 1.17 – Monte Carlo simulations: Depth dose curves for photons, protons and carbon ions. The curves represent the absorbed dose and do not take into account RBE values. The figure also shows a Spread Out Bragg Peak (SOBP) from protons covering a range of 3 cm with a homogeneous dose.

Figure 1.17 shows depth dose curves for photons, protons and carbon ions and illustrate the fundamental difference in dose deposition between photons and ions. The figure also shows how the position of the Bragg peak is a function of the primary beam energy. If the energy and the intensity of the beam are gradually varied, a series of overlapping Bragg peaks can add up to a homogenous dose deposited over the whole tumor or target volume. The sum of such overlapping Bragg peaks is known as the Spread Out Bragg Peak (SOBP). In Figure 1.17 a SOBP from protons covering the range from 5.6 cm to 8.6 cm is shown.
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In Figure 1.18 dose distributions from treatment plans for cancer in the central nervous system are shown. As part of this treatment the entire craniospinal axis is irradiated. The comparison of these plans illustrates how the sharp dose fall-off distal to the Bragg peak can be exploited in proton therapy to reduce the integral dose and spare critical organs, in this case the heart and lung.

![Dose distributions from treatment plans for cancer in the central nervous system](image)

*Figure 1.18: Example of dose distributions from standardized photon and proton treatment plans for cancer in the central nervous system. In the case of photon irradiation there is a significant amount of dose deposited in the exit region of the beam, while sparing of heart and lungs is better achieved using protons. Figure courtesy of Camilla H. Stokkevåg.*

As of today, the proton is the most commonly used particle for particle therapy purposes. However, carbon ion therapy has showed very promising results and the inherent features of heavy ions, such as their relatively high Relative Biologic Effectiveness (RBE), their ballistic properties and their effectiveness towards hypoxic cells, displays an even larger potential for the future. The particles found in the periodic system between the proton (hydrogen) and carbon ions are also candidates for radiation therapy, such as helium, lithium, and also oxygen.

There are two major methods of beam delivery in particle therapy. They are often referred to as active and passive delivery techniques. Ions and protons are charged particles and can thus be deflected, focused, and steered by magnetic fields. This gives the possibility to use a narrow beam for radiotherapy and steer the beam horizontally and vertically by the use of dipole magnets. The range of the particles
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can be adjusted by varying the energy of the beam. At GSI such a beam steering system known as the raster-scan technique was developed (Figure 1.19). In the treatment planning system, the target volume is dissected in layers in the longitudinal beam direction and each layer is covered in the beam orthogonal directions by a grid of pixels. A layer is irradiated pixel by pixel, and when the beam energy is changed, a new layer can be irradiated (Kraft, 2000b).

Figure 1.19: Intensity modulated ion therapy by the raster-scan principle developed at GSI. The target volume is divided into layers. Each layer is covered by a grid of pixels which are irradiated by a narrow pencil beam guided by dipole magnets. Modified from (Durante and Loeffler, 2010).

The passive beam delivery technique is described in Figure 1.20. A narrow particle beam is spread to a broad radiation field, either by introduction of material into the
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beam line or by magnets, which subsequently is shaped to fit the target volume. In addition, components such as ridge filters and energy degraders are used for energy adjustment and broadening. One of the concerns with the passive technique is the production of neutrons in the beam shaping components.

1.2.4 Secondary malignancies from radiation therapy

Radiation induced cancer has been recognized as a side effect of radiation therapy (van Leeuwen et al., 2000, de Vathaire et al., 1989). The risk from radiation is not only a function of dose but also varies with many factors including the size of the irradiated volume, fractionation scheme (dose rate), the degree of heterogeneity in the dose distribution as well as age and gender (BEIR, 2006, Bhatia and Sklar, 2002, Xu et al., 2008).

Most secondary tumors are found within the high dose region close to the target volume or in the primary radiation field, and it has been reported that only about 20% of all radiation-induced malignancies may be expected to develop far away from the treatment volume (Travis et al., 2002). These peripheral regions are exposed to lower dose levels originating mainly from scattered and secondary radiation, including a contribution from neutrons. Risk estimates of secondary cancers suffer in general from large uncertainties and lack of data. The most solid data concerning the carcinogenic effect of radiation comes from atomic bomb survivors. Models to estimate risk of secondary cancer from radiation therapy have been developed from radiation protection concepts (ICRP, 2007, BEIR, 2006), and risk factors are derived from atomic bomb survivors and extrapolated to low dose levels assuming a linear dose-response relationship. Effective doses are applied for risk estimation. These models have limitations and are the models are in general best suited to consider risks from low dose levels up to 1-2 Gy (Schneider, 2011). Such models may therefore be used for the low-dose regions where e.g. the neutron dose can be significant. However, for estimation of the total secondary cancer risk from radiation therapy,
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where it is necessary to include both the low and high dose areas, more elaborate models are needed.

The risk of secondary cancer from neutrons in passive proton therapy was estimated by Brenner and Hall (Brenner and Hall, 2008). The estimates are based on applying organ specific equivalent doses and risk factors from the BEIR-VII report (BEIR, 2006). In this case a 72 Gy lung treatment was given. The estimates indicated lifetime secondary cancer risks exceeding 10% for young females. The authors point out that the errors in such estimates are large, and as other authors have commented (Clasie et al., 2010), the RBE value of 25 for neutrons which is applied in these estimates may be too high.

![Figure 1.21: Proposed dose-response relationship for radiation induced carcinogenesis in humans (Hall, 2006).](image)

In conclusion it is apparent that the risk of radiation induced cancers is present in radiation therapy, and that the current risk models have large uncertainties. In Figure 1.21 a proposed dose-response curve for radiation induced cancer is presented (Hall, 2006). As the figure illustrates, the response to low doses is still far from fully understood, but the LNT model remains the most accepted model at low doses (BEIR, 2006).
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1.3 Monte Carlo simulations and treatment planning systems

1.3.1 Treatment planning systems

Dedicated planning software tools are used to obtain the best possible treatment configuration for each patient going through radiation treatment. Conventional treatment planning systems include algorithms such as the Pencil Beam or the Superposition/Collapsed Cone, and these are based on analytical models of radiation transport and their use requires extensive sets of measurements in order to verify the validity of the algorithms. These algorithms are very versatile due to their speed and calculation efficiency in conventional radiation therapy (Sikora, 2011). In particle therapy there is also the Relative Biological Effectiveness to consider. Especially for carbon ions the concept of biological dose optimization is a central part of treatment planning systems such as the TRiP98 developed at GSI (Krämer and Scholz, 2000).

The commercial treatment planning systems in clinical use today have known inaccuracies, in particular in the low dose areas including the normal tissue outside the treatment volume. Most treatment planning systems do not include a complete model for calculating the neutron dose that the patient will receive from the treatment. Studies have shown that dose is underestimated in areas which, according to the treatment planning system, receive less than 5% of the dose delivered to the target volume (Howell et al., 2010).

1.3.2 Monte Carlo simulations

Monte Carlo simulations represent a valuable tool to theoretically predict the outcome of specific experiments through advanced computer modeling. A Monte Carlo simulation relies on repeated random sampling from probability distributions representing the possible physical interactions relevant for the simulation. Such an approach is well suited for modeling of radiation transport through complex geometries. In this context single particle transport and interaction histories are
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recorded. A high number of independently simulated histories are added up to represent a numerical solution of the problem on the basis of the defined geometry for the simulation and the cross sections for the relevant interactions.

Traditionally, Monte Carlo simulation has been a research tool and not for use in the clinics. One of several reasons for this is that the computational power needed has resulted in a calculation time not acceptable for daily treatment planning in the clinics. Contributions to errors in Monte Carlo simulations include uncertainties in the geometrical definitions of the experiments and uncertainties in the applied interaction cross sections. The simulation of neutron production is in particular challenging and the agreement in neutron dose simulations is typically not at the same confidence level as when simulating dose from protons, photons or electrons (Paganetti, 2012). Still, Monte Carlo simulations enable the possibility to study low level and peripheral doses in a detailed way which is not possible using traditional treatment planning software. This is necessary to do, and in this respect Monte Carlo simulations are invaluable in the context of investigating neutron doses and the risk for radiation induced secondary cancer.
2. Experimental set-up

The objective of the experimental work in this thesis was to measure the thermal and fast neutron fluence in irradiation experiments with photons, protons and carbon ions.

The majority of the experiments was based on measurements around single fields or around pencil beams and implied the use of water phantoms. A simple set-up was chosen to provide a baseline for a systematic overview of the neutron dose produced in radiation therapy, independent of the local flavor or design of treatment plans. A simple set-up also enabled the option of conducting Monte Carlo simulations for verification of the experimental results.

2.1 Detectors

In the radiation field from a medical linear accelerator, the accelerator delivers bursts of radiation in short, intense pulses. For this reason, a detector which is not sensitive to photons, and which can be used without dead time (pile up) problems due to high intensity in the pulses, would be preferable. In measurements of neutron dose produced during particle therapy the demands to the detector are different because the neutron energy extends up to several hundreds of MeV and charged particles such as protons and heavier ions are present in the radiation field.

In this work SRAM based neutron detectors, bubble detectors and thermoluminescence detectors (TLDs) have been the main tools for neutron detection. The SRAM detectors, covering a wide energy range, were applied in proton and carbon ion beams, while the bubble detectors were used to measure neutrons from photon therapy due to their high sensitivity to neutrons and because they show little response to gamma radiation. Thermoluminescence detectors were applied in photon therapy for assessment of the thermal neutron dose. In addition to this, diodes and ionization chambers were used for measuring the photon field from the medical linac used in this work.
A number of different neutron detectors are commercially available today. Passive detectors are widely used and have the advantage of being compact and relatively cheap, but they often lack the ability to give information about energy or discriminate against other radiation than neutron radiation. In addition, passive detectors may be cumbersome to work with, and if beam time is limited active detectors may help in a better utilization of the available timeslot for measurements at busy facilities.

Active (electronic) detectors also come in many forms. Gas filled proportional counters are widely used neutron detectors. $^{3}$He-gas and BF$_3$-gas are commonly used. Boron and Helium-3 nuclei have high cross sections for thermal neutrons and detectors are often designed with moderating material surrounding the gas detector volume in order to increase the detector response to fast neutrons. Unfortunately, the detectors thus become large, and are not suited for measurements with high spatial resolution, e.g. inside phantoms or close to other equipment.

There is still abundant room for further improvements of the equipment in the field of neutron detection.

**2.1.1 The SRAM (Static Random Access Memory) detector**

A neutron detector based on radiation effects in SRAM (Static Random Access Memory) chips has been developed at the University of Bergen for this project (Velure, 2011, Larsen, 2011). Testing and characterizing the device has been a part of this present Ph.D. work. The motivation for the use of SRAM based detectors in particle therapy is the need for detectors which are active, inexpensive, small in size and insensitive to photons. The compact detector design enables in-phantom measurements with sub-cm spatial resolution.

The detector is insensitive to gamma radiation and counts so-called soft errors induced by single high energy particles traversing the detector. A soft error is an effect which does not give permanent damage, but may reversibly alter the state of the device.
2 Experimental set-up

The SRAM detectors are based on four identical 16 Mbit SRAM memory chips from the vendor Cypress. Each SRAM chip consists of $16 \times 2^{20}$ individual memory cells or bits. Further technical information can be found in Table A-1 in the appendix. The SRAM detector is sensitive to neutrons with energies down to about 3 MeV.

Traversing particles may undergo inelastic collisions, mainly with silicon in the SRAM, producing high LET charged fragments. These secondary particles can deposit enough charge in a memory cell to change the logic state of this specific cell. This process is referred to as a bit flip, and when such a process occur from interactions of one incoming nuclear particle it is referred to as a single event upset (SEU). The rate of which the SEUs occur is then proportional to the high energy particle flux.

The common single event effects include:

- Single Event Upset (SEU): The stored data value of a bit is reversed.
- Multiple Bit Upset (MBU): Several bits are reversed by a single event.
- Single event latch up (SEL): A low resistance path between ground and power supply occur from a spark caused by one particle. The current state will remain after the triggering event is removed, and remain until power is removed or a severe failure occurs (Sexton, 2003).

The principle of detecting SEUs follows three basic steps:

1. Write a known pattern to the SRAM (e.g. 101010101010...)
2. Read the SRAM continuously while it is exposed to radiation and check if the pattern has changed.
3. If a change in the data is detected, a SEU has occurred and will be registered.

In this way, the count rate is proportional to the flux of particles capable of inducing SEUs. The detector’s neutron cross section is defined as the number of registered SEUs per neutron fluence ($\text{SEU/(n cm}^{-2}$)).

The SRAM is sensitive to protons, neutrons and heavier particles. Due to the Coulomb barrier, the probability of inducing SEUs is higher for neutrons than for protons at low
2 Experimental set-up

energies. At energies above 50 MeV, the Coulomb barrier becomes less important and the inelastic cross section is similar for protons and neutrons. Consequently, the ability to induce SEUs is expected to be similar above 50 MeV.

At low energies, the neutron interaction that is most likely to cause a SEU in the detector is the $^{28}$Si(n,$\alpha^{25}$Mg reaction (Larsen, 2011). The cross section for the $^{28}$Si(n,$\alpha^{25}$Mg reaction can be seen in Figure 2.1 and indicates a lower energy threshold of 2-3 MeV for SEU induction through this reaction.

Thermal neutrons may also cause SEUs through neutron capture reactions. This is especially the case for SRAM chips which contain boron as a dopant or has this material in one of the insulating layers. However it is difficult to know if an SRAM chip contains boron or not, and it may then be needed to irradiate SRAM chips in a thermal neutron field to determine if they are sensitive to thermal neutrons or not.

Figure 2.1: The cross section for the $^{28}$Si(n,$\alpha^{25}$Mg reaction. This represents an important interaction for single event upset induction in the SRAM detectors. Data from (Chadwick et al., 2006).

Ions heavier than the proton, such as deuterons, tritons, alpha particles and heavier fragments may also induce SEUs. The SEU cross sections for these particles need further investigation. More details on Single Event Upsets in SRAMs can be found in
Experimental set-up

(Røed, 2009) and (Larsen, 2011). The detector hardware, firmware, and software is further described in (Velure, 2011).

A picture of the SRAM detector is shown in Figure 2.2. The sensitive area of the SRAM chip is the die (marked with a red rectangle). Due to the die’s small dimensions (6.3 x 7.8 mm²) the spatial resolution achievable is quite good for most purposes in neutron dosimetry. The best spatial resolution is achieved by treating each SRAM chip as an individual detector.

![Figure 2.2: A picture of the SRAM detector showing the four SRAM chips to the right. The red rectangle marked on one SRAM indicates the size of the radiation sensitive area for each chip.](image)

2.1.2 Bubble detectors

When a liquid continues to exist in the liquid state above its normal boiling point, it is said to be superheated. Boiling or nucleation can be retarded until the temperature of the liquid reaches its so-called superheated limit (Eberhart et al., 1975).

Bubble detectors are based on superheated liquids and use a similar principle as the bubble chamber. The detectors are primarily used to detect neutrons. The bubble detectors use superheated droplets dispersed throughout a clear elastic polymer. Neutrons passing through the detector may produce small visible bubbles, and the polymer ensures that the bubbles are trapped at the sites of formation (Andrews et al., 2006). Based on the number of bubbles formed in the detector it is possible to
calculate the neutron fluence and dose. An automatic bubble counter from BTI was used for bubble counting in this work enabling a reproducible counting procedure.

The bubble detectors show little response to gamma radiation, and they are passive detectors. These two properties are very important as they enable the possibility to measure neutrons in areas with significant gamma background and in pulsed radiation fields.

A fraction of the neutrons which traverse the bubble detector will interact with the detector. These interactions give rise to a variety of secondary charged particles, including recoil ions (Ing et al., 1997). The charged particles will slow down in accordance with the stopping power of the ions in the medium at the interaction site. Some of the secondary charged particles will deposit energy in the droplet. The passage of the particle within the superheated droplet will give rise to a trail of microscopic bubbles. The bubbles that exceed a critical bubble size, $r_c$, will grow into macroscopic bubbles, while bubbles of smaller radii will collapse from the effects of surface tension (see Figure 2.3).

The vendor Bubble Technology Industries (BTI) has developed the Bubble Detector Spectrometer (BDS) consisting of six different types of bubble detectors, each with its own detection energy threshold for neutrons. This enables the possibility of obtaining
2 Experimental set-up

crude neutron spectra in demanding mixed radiation fields. The response of the detectors to neutrons as a function of energy is shown in Figure 2.4. The neutron cross section for the bubble detector is defined in the units bubbles per neutron fluence (b/(n cm\(^{-2}\)).

![Response functions for the 6 different BDS bubble detectors (BTI, 2009).](image)

Figure 2.4: Response functions for the 6 different BDS bubble detectors (BTI, 2009).

BTI has also developed the Bubble Detector Thermal (BDT) for detection of thermal neutrons. Lithium (\(^{6}\)Li) is dispersed throughout the detectors polymer gel, and thermal neutrons are captured through the reaction \(^{6}\)Li(n,\(\alpha\))T. The reaction releases an amount of energy which may be enough to expand one of the microscopic droplets in the detector into a macroscopic bubble. Fast neutrons may also create macroscopic bubbles in the detector through elastic nuclear collisions producing energetic recoil ions.

**The BUNTO code**
The BUNTO code (Ongaro et al., 2001) is especially developed to unfold the results from passive neutron detectors including the BDS detectors. The motivation for applying BUNTO was to improve the accuracy of the spectral unfolding of the bubble
2 Experimental set-up

Figure 2.5: The pictures show the BDS (white cap) and BDT detectors. To the left a picture of an irradiated detector is seen and the bubbles formed by neutrons are visible.

detector measurements. The simple unfolding procedure provided by the BDS vendor is based on so called spectral striping where error is known to accumulate leading to large errors in the lower energy region of the spectrum. BUNTO uses an iterative method to calculate the spectrum. It is based on random sampling of unfolding data from a normal distribution, where the parameters of the distribution (the standard deviation and mean value) are the statistical error and mean experimental values from the measurements (Zanini et al., 2005). The final spectrum obtained from the BUNTO code is the mean of possible solutions of the unfolding, weighted on the mean standard deviation. On request from the user, BUNTO limits the maximum difference between possible solutions and the mean value to 20%. The BUNTO code is further described and applied in (Ongaro et al., 2000, Ongaro et al., 2001, Zanini et al., 2005).

2.1.3 Thermoluminescence detectors

Thermoluminescence detectors (TLDs) are widely used in dosimetry. TLDs are passive detectors found in a variety of shapes, and can be made from different materials depending on which type of radiation they are intended to detect.

To understand the mechanisms involved in the process of thermoluminescence, some basic understanding of solid state physics is needed. In an individual atom, electrons occupy discrete energy levels. In a crystal lattice, on the other hand, electron energy levels are perturbed by mutual interactions between atoms and give rise to energy levels referred to as “allowed” and “forbidden” energy bands. In addition, the presence
of impurities in the crystal creates energy traps in the forbidden region, providing metastable states for the electrons (Khan, 1994). When the material is irradiated, some of the electrons in the valence band (ground state) receive sufficient energy to be raised to the conduction band. The vacancy created in the valence band is called a positive hole. The electron and the hole move independently through their respective bands until they recombine (electron returning to the ground state), or until they fall into a trap. This is illustrated in Figure 2.6.

![Figure 2.6](image)

*Figure 2.6: The figure shows a simplified illustration of thermoluminescence. Modified from (Khan, 1994).*

If there is instantaneous emission of light owing to these transitions, the phenomenon is called fluorescence. If an electron in the trap requires energy to get out of the trap and fall to the valence band, the emission of light in this case is called phosphorescence (delayed fluorescence). If phosphorescence at room temperature is very slow, but can be speeded up significantly through heating, the phenomenon is called thermoluminescence.

**TLD neutron dosimetry**

There are no neutron sensitive TL elements available which are not also sensitive to photons. In order to obtain neutron doses it is therefore necessary to use two types of TL elements, one which is sensitive to both neutrons and photons and one which is only sensitive to photons. Then, the photon dose can be subtracted, and the dose from neutrons can be obtained. The TLD-600 and TLD-700 are common TL elements used for neutron dosimetry. The TLDs are both based on LiF:Mg,Ti, lithium fluoride with added magnesium and titanium.
2 Experimental set-up

The TLD-600 is enriched with $^6\text{Li}$ resulting in a large cross section for thermal neutrons while in TLD-700 $^6\text{Li}$ is separated out and 99.93% of the lithium is $^7\text{Li}$, resulting in a crystal relatively insensitive to thermal neutrons in comparison. The photon sensitivity of the two crystals is similar. By measuring the photon and neutron sensitivity of the TLD-600 and the photon sensitivity of the TLD-700 the thermal neutron dose from a measurement using both detectors can finally be calculated as shown in the following equations (Howell et al., 2005):

$$R_n = R_{600}^{n+\gamma} - \frac{R_{700}}{k}, \quad (2.1)$$

where $k$ is the relative response to photons for the two TL elements:

$$k = \frac{R_{700}}{R_{600}^{\gamma}}. \quad (2.2)$$

$R_n$ is the TLD-600's response to neutrons, $R_{600}^{n+\gamma}$ is the total response of the TLD-600 element, $R_{700}$ is the response from the TLD-700 and $k$ is the ratio of the gamma sensitivities of the two TLD types. The neutron sensitivity of TLD 700 is assumed to be negligible and therefore $R_{700}^{\gamma}$ is equal to $R_{700}$.

The light emitted from the TLDs is transformed to a current by a photomultiplier, and the total charge output, usually given in nano Coulombs (nC), is given by the TLD reader. The charge can be converted into dose based on calibration of the TL elements in known radiation fields. For neutron dosimetry, the TL elements must be exposed to a thermalized neutron beam or a neutron source with a known fluence.

The read out process includes three steps. Initially a preheating of the detectors is applied to segregate the light generated from low-energy traps to minimize fade effect. Then the data collection starts in the acquire phase. Following the recommended acquire settings from the vendor means a read out with increasing temperature up to 300 - 400 degrees depending on the TL type. Finally the crystals are annealed at high
2 Experimental set-up

temperature to ensure that prior exposures will not affect future measurements with the detectors.

TLDs may be applied also in particle therapy, but should be used with caution. While in photon therapy the ionization density in the irradiated volume is in general homogenous, the ionization in particle therapy is characterized by an inhomogeneous distribution. The efficiency of the TLDs is also dependent on the LET and particle species (Berger and Hajek, 2008). The energy deposition from charged particles in the crystal may be in the sublinear, linear or superlinear dose response range for the TLD. The relative efficiency for charged particles compared to photons may therefore be more than 1, but also less than 1. This complicates the interpretation of the measurements with TLDs in particle therapy. More on TLDs in particle therapy can be found in (Berger and Hajek, 2008).

2.2 Experimental facilities

The pursuit of the objectives in this thesis led to experimental work at a number of irradiation facilities with predetermined beams. This included irradiations in proton beams, carbon ion beams, mixed hadron radiation fields, thermal neutron fields and quasi-monoenergetic neutron beams. Quasi-monoenergetic neutrons beams are characterized by a peak in the beam energy distribution similar to monoenergetic sources, but quasi-monoenergetic neutron beams also comprise a low-energy tail of neutrons that extend below the peak energy.

A custom made water phantom with special holders for the bubble detectors, the SRAM detector, and the TLDs was used for the neutron fluence measurements. The dimensions of the phantom was 12 x 18 x 23 cm³ (W x L x H). The phantom was made from 5 mm thick PMMA walls with density 1.20 g/cm³ while the detector holders were made from polycarbonate (Lexan) with density 1.20 g/cm³. Water phantoms have previously been shown to be suited for the purpose of investigating neutron doses from particle therapy (Halg et al., 2011).
2 Experimental set-up

At an early stage in the project, a number of different SRAM detectors were tested for suitability as a detector for neutrons in radiation therapy. After the initial experiments, the final detector calibration in the Uppsala neutron beam was performed only with the detector found to be best suited, the 16 Mbit SRAM detector presented in section 2.1.1.

2.2.1 Physikalisch - Technische Bundesanstalt (PTB)

The PTB in Braunschweig offer quasi-monoenergetic neutron beams of several energies and gives the opportunity to test neutron detectors in a predetermined radiation field.

The objective of the measurements at PTB was to calibrate the SRAM detectors in the fast neutron beams available at PTB. Measurements were performed in neutron beams with energies of 5.8 MeV, 8.5 MeV, and 14.8 MeV. Figure 2.7 illustrates the deuterium beam hitting the target producing the neutron beam which then reaches the detectors placed at 1 meters distance.

![Diagram of set-up at PTB](image)

*Figure 2.7: A sketch of the set-up at PTB. A deuterium beam on a deuterium gas target was used for the 5.8 MeV and 8.5 MeV beams, while a tritium target was used to produce the 14.8 MeV beam.*

**BDT Fast neutron calibration**

When measuring thermal neutron doses, fast neutrons may also be present in the radiation field. These fast neutrons may then cause an overestimation of the thermal neutron dose if they contribute to the detector response. The objective of the fast neutron measurements was to measure the BDT bubble detectors sensitivity to fast
Experimental set-up

neutrons in order to see if fast neutrons may have an impact on the results during measurements of dose from thermal neutrons with BDT detectors.

Measurements were performed in neutron beams with energies of 5.8 MeV, 8.5 MeV and 14.8 MeV, and the set-up is illustrated in Figure 2.7.

2.2.2 Oslo Cyclotron Laboratory (OCL)

The Oslo Cyclotron Laboratory is a part of the center for accelerator based research and energy physics (SAFE) at the University of Oslo. The cyclotron at OCL can accelerate protons, deuterons, $^3$He and $\alpha$-particles. The OCL offers proton irradiation for detector testing in the energy range 2 – 35 MeV.

At the Oslo cyclotron laboratory a number of different SRAM chips were tested and calibrated in a 26 MeV proton beam. Figure 2.8 shows the setup with a SRAM detector placed in the measurement position (Larsen, 2011).

![Figure 2.8: Set-up at the Oslo Cyclotron Laboratory. The SRAM detectors were irradiated with 26 MeV protons.](image)

2.2.3 CERN-EU Reference Field (CERF)

The CERF facility is installed in one of the secondary beam lines from the Super Proton Synchrotron (SPS). A positive hadron beam of 60.7% protons, 34.8 % pions and 4.5 % kaons, with 120 GeV/c momentum collides with a copper target creating a mixed hadron radiation field covering a wide energy range (Larsen, 2011). The
response from this radiation field expected to give a good measure of the SRAM detectors plateau sensitivity to neutrons and other hadrons. The intensity of incoming particles was monitored by a precision ionization counter (PIC).

**Figure 2.9:** Positioning of the SRAM detectors during one of the measurements in the mixed hadron radiation field at the Cern-EU reference field (CERF) (Larsen, 2011).

### 2.2.4 Institute for Energy Technology (IFE), Kjeller

The Institute for Energy Technology (IFE) at Kjeller, Norway, administers the Joint Establishment Experimental Pile (JEEP II) nuclear research reactor. The reactor uses low enriched $^{235}$U as fuel. Heavy water in the reactor core provides both cooling and moderation of neutrons. A part of the neutrons escaping the reactor core are further moderated in a large water pool. The radiation field in the outer pool, dominated by thermal neutrons, is used for detector testing. The thermal neutron fluence in the measurement positions was measured through gold activation measurements conducted by the IFE staff.

**SRAM thermal neutron calibration**

The irradiation of the SRAMs was performed inside an aluminum pipe in the left water pool as shown in Figure 2.10. The device under test was lowered down to the bottom of the pipe corresponding to the measurement position indicated in the figure.
2 Experimental set-up

**Figure 2.10:** The sketch shows the positions for thermal neutron measurement at the JEEP II reactor at Kjeller. The detectors were lowered into a pool of water for irradiation a predetermined length of time.

**BDT and TLD Thermal neutron calibration**

The detectors were placed inside a lead container, reducing the intensity of the photon background, and lowered down into the water pool of moderating water as seen in Figure 2.10. The lead ensured that the background signal from photons was decreased significantly, and thus enabled a more precise calibration of the TLD-600 detectors response to thermal neutrons. The exposure times for the measurements were four, eight, and thirty minutes. For the four minute exposure twelve crystals of each type were irradiated while for the longer runs, six crystals of each type was used. Four different BDT bubble detectors were repeatedly irradiated four times. Each exposure was approximately two minutes.

**2.2.5 The Svedberg Laboratory (TSL), Uppsala**

The Svedberg Laboratory offers both neutron and proton beams to its customers. The Gustaf Werner Cyclotron accelerates protons to a standard energy of 180 MeV (TSL, 2013). TSL has a long history of proton therapy and the first proton treatment in the world was delivered there in November 1957. Currently patients which are treated at TSL with proton therapy alone or in combination with photon therapy include patients with metastases to the brain, prostate cancer, iris melanomas, children with brain
tumors, chordomas and chondrosarcomas at the base of skull, certain other head and neck cancers and malignant gliomas.

**Neutron beam irradiation**

At TSL the SRAM detector was calibrated in a quasi-monoenergetic neutron beam. The beam was produced by a 180 MeV proton beam from the TSL cyclotron incident on a lithium-7 target of 23.5 mm thickness, as shown in Figure 2.11. After the target a bending magnet deflects the charged particles from the forward path, leaving a neutron beam with an average energy of 174.5 MeV for the peak neutrons. The full spectrum can be seen in Figure 2.11.

![Neutron beam diagram](image)

Figure 2.11: The neutron beam was produced by a 180 MeV proton beam incident on a lithium target. The neutron fluence plateau region was 17.6 cm in diameter at the measurement position. The neutron spectrum produced from the proton beam is shown in the plot to the right in the figure. The average peak energy was 174.5 MeV, and the spectrum was calculated by the TSL staff with the algorithm described in (Prokofiev et al., 2002).

Three separate monitors were used for measuring the beam fluence in the neutron experiment. The relatively large beam diameter of 17.6 cm enabled the option of measuring with several detectors simultaneously. In Figure 2.12 two pictures of the SRAM detectors in the beam are shown. Four detectors were placed inside the beam simultaneously. The red circle in the figure illustrates the neutron beam diameter at the measurement position.
Experimental set-up

Figure 2.12: The pictures show the SRAM detectors in position for neutron irradiation at TSL. To the left the neutron collimator is also seen. The four SRAM chips on each detector can be seen in the center of the picture to the right. The red circle indicates the diameter of the neutron beam.

Proton beam irradiation

The proton beam irradiation took place at the PAULA facility at TSL. The proton beam from the cyclotron was used directly without collimators or degraders. This ensured that the only significant source of neutrons was the water target. In this way the experiment represented a situation similar to a scanning beam facility. The proton beam energy at the user position was approximately 178 MeV. During the experiment, the proton beam profile was measured by the TSL staff to be Gaussian with FWHM 1.33 cm ($\sigma = 5.6$ mm).

The SRAM detectors were placed in a custom made water proof holder inside the water phantom. In addition, SRAM detectors were positioned outside the phantom for measurement of the angular distribution of the neutrons. For the Bragg peak to be at the desired position, several thin walled polystyrene flasks with water were placed in front of the larger water phantom. The flask dimensions were 4 x 12.2 x 20 cm$^3$. A sketch of the set-up can be seen in Figure 2.13, while Figure 2.14 show pictures of the set-up. At the beginning and end of the measurements sessions Gaf-chromic film irradiations confirmed the position of the proton beam. The measurements at 21.2 cm
2 Experimental set-up

depth are assumed to represent the fluence at Bragg peak depth although the Bragg peak depth is approximately 21.0 cm.

Figure 2.13: A sketch of the set-up in the proton beam at TSL. For the lateral fluence measurements, the SRAM detectors were placed at 7 different distances from the beam axis. In addition angular measurements as well as measurements at different depths inside the phantom were performed.

Figure 2.14: Set-up for SRAM detectors in the proton beam. Measurements are performed inside the phantom and at 0°.

2.2.6 GSI Helmholtz Centre for Heavy Ion Research

GSI is a facility for research in fundamental physics and especially heavy ion research. Since 1997, GSI has been operating a radiotherapy unit for cancer treatment using
Experimental set-up

carbon ions. Today, more than 400 patients have been treated at GSI for tumors in the head and neck region. Also patients with tumors along the spinal cord and prostate cancer have been treated at GSI. Knowledge and experience has now been transferred from the research facilities at GSI to the clinic at Heidelberg ion-beam (HIT) radiation therapy center.

**Measurements with a water phantom**

A water phantom was irradiated by a 200 MeV/u carbon ion beam at GSI. The beam was applied directly without collimators or degraders. For this beam energy the Bragg peak is situated at 8.6 cm depth in water. At this depth, measurements with the SRAM detector were performed at different lateral distances. Measurements were also performed at various angles with a distance of 15 cm to the central Bragg peak position. The set-up is illustrated in Figure 2.15 and Figure 2.16. An ionization chamber was placed in front of the phantom to monitor the beam intensity. The phantom consisted of three thin walled polystyrene flasks filled with water, and the total dimensions of the water phantom was 12.0 x 12.2 x 20.0 cm³ (D x W x H).

![Figure 2.15: Set-up for the water phantom measurements with the SRAM detector with the 200 MeV/u carbon ion beam at GSI. To the left the angular measurement positions are shown while the figure to the right shows the principle of the lateral fluence measurements. The ionization chamber is denoted IC in the figure.](image)

**Measurements with an antropomorphic phantom**

Neutron fluence measurements were performed at GSI with the SRAM detector behind an anthropomorphic phantom irradiated with a dose planned carbon ion
2 Experimental set-up

![Image of set-up](image)

**Figure 2.16**: A picture of the set-up for the carbon beam irradiation at GSI. In this case the detectors are placed at 0° and 30°. The ionization chamber and beam exit window is seen to the right.

The purpose of this set-up was twofold; to measure the neutron dose in the depth beyond the target volume in a typical head treatment, and to investigate the option of discriminating signal from charged particles in the SRAM detector by the use of scintillators. The detector was placed at zero degrees behind the phantom at 12 cm distance. A sketch and a picture of the set-up can be seen in Figure 2.17.

![Sketch of set-up](image)

**Figure 2.17**: Set-up of the measurements in the carbon ion therapy room at GSI. The SRAM detector is mounted between two scintillators to the left in the picture, while the beam exit window is seen to the right.

The target volume was 5 x 5 x 2.5 cm³, and the treatment plan required beam energies from 160 MeV/u to 270 MeV/u. Further, 4.32 x 10⁸ carbon ions were needed to deliver
Experimental set-up

1 Gy to the target volume. The phantom had a density of 0.997 g/cm$^3$ and was composed of 34.7% carbon, 56.1% hydrogen, 1.1% nitrogen and 0.01% antimony.

Two scintillators of the type BC-400 were used to measure the amount of charged particles. These were placed on both sides of the SRAM detector and a coincidence signal from the scintillators gave the number of charged particles passing through the SRAM. The overlapping area covered by both scintillators was estimated to be 7 x 8 cm$^2$.

2.2.7 Haukeland University Hospital

At Haukeland University Hospital (HUH) several measurement series were conducted to investigate the neutron and photon doses outside the treatment field in radiation therapy with photons. A Varian Clinac 23 iX medical linear accelerator in the 15 MV energy mode was used in the experiments. The BDS bubble detectors were used to measure the fluence of neutrons above 10 keV, while BDT bubble detector and TLD-600/700 crystals were applied to measure thermal neutrons. Furthermore, FLUKA Monte Carlo simulations based on a geometrical model of the accelerator were conducted for comparison with the measurements.

**Neutron fluence measurements**

Measurements of neutron fluence were performed inside and outside the custom made water phantom. The detector response varies with temperature and the water temperature was therefore controlled and for measurements outside the phantom the detectors were placed in insulating Styrofoam holders.

The neutron fluence measurements were conducted using a 3 x 3 cm$^2$ photon field shaped by the jaw collimators. The MLCs were retracted during these measurements. The BDS, BDT and TLD detectors were positioned as described in Figure 2.18 and Figure 2.19.

The measurements of neutron fluence and neutron energy spectra with the BDS detectors include irradiation of detectors with 6 different thresholds. For each detector
2 Experimental set-up

**Figure 2.18:** The figure shows a sketch of the experimental set-up for the neutron fluence measurements at HUH. A 3 x 3 cm\(^2\) photon field was applied in the 15 MV energy mode.

**Figure 2.19:** A picture of the experimental set-up with the BDT detectors and a 3 x 3 cm\(^2\) photon field at HUH. Three bubble detectors are placed inside the water phantom for irradiation.

...type, 6 individual detectors were irradiated for each measurement position. The exception was for the BDS 10 000 which did not show any response even at high doses and the BDS 10 000 detectors were therefore not repeatedly irradiated. In total, more than 30 detectors were irradiated for each measurement position to ensure good statistics adding to the reliability of the results.
Experimental set-up

The unfolding of the spectra with the BUNTO code was based on response from all of the six detector types as described in section 2.1.2. No initial guess spectrum was introduced, and the final spectra are therefore based on the experimental data alone.

The thermal neutron measurements were repeated with three different BDT detectors and four TLDs (two TLD-600 and two TLD-700) in each measurement position.

2.3 Monte carlo simulations

The Monte Carlo simulations in this work have been performed with the FLUKA Monte Carlo tool (Battistoni et al., 2007, Fasso et al., 2005). FLUKA is a general purpose Monte Carlo particle transport code covering a wide range of applications including basic research in high energy physics, radiation protection, dosimetry and a number of other applications. The use of FLUKA in Monte Carlo simulations related to radiation therapy, including protons and carbon ions, has previously been evaluated (Parodi et al., 2012, Sommerer et al., 2006).

In FLUKA, neutron cross sections are derived from the most recently evaluated data (Fasso et al., 2011). The neutron cross sections are continuously enriched and updated on the basis of the most recent evaluations (e.g. ENDF/B, JEF, and JENDL). Neutrons of energies below 20 MeV are treated in a group wise manner in FLUKA with a dedicated neutron cross section library consisting of 260 neutron energy groups.

Nuclear interactions generated by ions are in FLUKA handled through interfaces to external event generators. In the energy range up to 125 MeV per nucleon, a model based on the Boltzmann master equation theory (Cerutti et al., 2006) is applied. At higher energies (up to 5 GeV/n) a Relativistic Quantum Molecular Dynamics model (RQMD) is employed (Sorge et al., 1989).

The Monte Carlo simulations were performed using a cluster of Linux machines at Uni. Computing located in Bergen. Typically 100 independent simulations, each with
2 Experimental set-up

a high number of primary particles, were performed for each Monte Carlo simulation set-up. The statistical errors were for the majority of the results below 2%.

2.3.1 Proton beams

The energy information and the beam profile measured at The Svedberg Laboratory in Uppsala were used as input for FLUKA Monte Carlo simulations of the proton beam experiment. Neutron fluence estimates and neutron energy spectra have been obtained for all the fluence measurement positions. Because the biological effect of neutrons is strongly dependent on neutron energy, this information is important for the dose estimates.

In addition, Monte Carlo simulations were performed with a proton beam of 107 MeV. These simulations enable a comparison to measurements and simulations with 200 MeV/u carbon ions because the range is the same for these two beams.

2.3.2 Carbon ion beams

The Monte Carlo simulations of neutrons from carbon ions include simulations of the same set-up as the proton beam experiments, but with carbon ion beams of 200 MeV/u and 341 MeV/u with ranges equivalent to the 107 MeV and 178 MeV proton beams.

Monte Carlo simulations were also performed to reproduce the experimental conditions during the SRAM measurements in the carbon ion beams at GSI. The 200 MeV/u carbon ion beam experiment was simulated in order to compare with measurements. Furthermore, a Monte Carlo simulation based on the delivered treatment plan at GSI was designed in an attempt to reproduce the experimental conditions from the irradiation of the anthropomorphic phantom. The phantom shape was simplified to comprise a cylinder with the approximate dimensions of the phantom head and the ICRU defined soft tissue material was assigned to the phantom. The Monte Carlo simulation does not take into account details such as ripple filters and fragmentation in bone.
2 Experimental set-up

2.3.3 Photon beams

**Geometry definition**
When simulating neutron production in photon therapy one has to consider a specific medical linear accelerator and define the geometric model of the Monte Carlo simulation based on this. In this work the treatment machine is the Varian Clinac 23 iX accelerator. The geometrical definitions of the major components are based on the technical drawings from the vendor.

**Target**
The target is made from tungsten and copper. The primary electron beam produces bremsstrahlung in the tungsten, while the copper is primarily present to ensure fast heat dissipation.

**Primary collimator**
Surrounding the target is the primary collimator. This is made from tungsten due to this material’s ability to effectively shield photons and electrons. With a conical cavity inside, it is the first photon beam shaping component. At the lower end of the primary collimator the vacuum window is mounted. This is made from beryllium and it ensures that the photon beam leaves the vacuum and enters the air in the treatment room with minimal perturbation.

**Flattening filter**
The purpose of the flattening filter is twofold. It will, due to its conical shape, flatten the beam ensuring a more homogenous dose distribution and in addition harden the photon beam. With intensity modulated beams the flattening filter becomes less important and the use of beams which are not flattened may become more common.

**Jaws (secondary collimators)**
The jaws are tungsten blocks which are used for field shaping. They are sometimes referred to as the X and Y collimators as they define the photon field size in the x and y direction.
2 Experimental set-up

**Multi leaf collimators**
The final beam shaping components before the beam reaches the patient are the multi leaf collimators (MLCs). Tungsten is the main component also in the MLCs. The MLCs can produce complex field shapes and dynamic fields. Because no dynamic treatments have been simulated in this work, the MLCs are geometrically defined as two blocks instead of the individual leaves.

**Other components**
Several additional components have been simulated such as the ionization chambers for beam monitoring, the treatment room with concrete walls and the phantoms used for the specific experiments. Additional shielding components and electronics in the linac head has not been modeled due to the lack of geometrical information on these components.

**Biasing**
Importance biasing has been used to achieve sufficient statistics in the Monte Carlo simulation of neutrons from photon therapy. This concept implies that the geometry is divided into regions of different importance with respect to the neutron fluence needed in those regions to get reliable results. A neutron entering a region with importance value 5 times higher than the importance of the previous region, will be split into 5 neutrons with the same properties as the original. When used with caution this is a powerful tool to improve the statistics in a Monte Carlo simulation.

The LAM-BIAS option in FLUKA was applied to improve the statistics of photonuclear reactions which, compared with electromagnetic photon reactions, has relatively low cross sections. In addition, the photon and electron transport thresholds are set to 6 MeV to save computing time in the neutron Monte Carlo simulations. Because the neutron energy separation energy is found at higher energies this should not have an impact on the results from the Monte Carlo simulation.
Experimental set-up

**Verification of the Monte Carlo linac model**

Measurements of photon dose profiles and depth dose curves were performed at the linac to confirm the validity of the Monte Carlo model. The measurements were performed in a water phantom with ionization chambers from PTW as both field and reference detectors.

In order to compare not only relative measurements and Monte Carlo simulations, but absolute dose it is necessary to investigate, through Monte Carlo simulations, the amount of primary electrons needed to give a certain dose in the isocenter. The neutron fluence and dose is recorded per primary electron in the simulations, and when converting the fluence into neutron dose per treatment Gy or per Monitor Unit (MU) the number of primaries needed to deliver one Gy or one MU must be known. Monte Carlo simulations were performed with a spherical detection volume at the isocenter. The settings for this simulation included 10 x 10 cm² photon field size, the 15 MV energy mode, Source to Surface Distance (SSD) of 90 cm and isocenter positioned at 10 cm depth in water.

**The origin of neutron production**

Monte Carlo simulations were performed to investigate where neutrons are created, how they propagate and where they deposit dose. Two separate Monte Carlo simulation methods were used for this purpose. The first method was to investigate the neutron yield of major linac components. The second approach was based on detecting neutrons depositing dose in a specific part of the water phantom and trace the origin of these neutrons back to the component where they were created. The detection regions are referred to as *trace regions*. This latter approach gives specific information on to which degree different components contribute to neutron dose in the water phantom.

Because the results may depend on the settings of the collimators, four different configurations have been simulated and an overview of these is given in Table 2.1. In all these cases the beam quality was 15 MV, and trace regions both inside and outside the photon field were considered as seen in Figure 2.20.
Table 2.1: Settings for the neutron production and origin trace Monte Carlo simulations. MLCs set to 10 cm indicate that the MLCs produce a photon field of 10 cm width in the cross plane direction.

<table>
<thead>
<tr>
<th>Collimator setting no.</th>
<th>Jaws</th>
<th>MLCs</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>3 x 3 cm$^2$</td>
<td>Retracted</td>
</tr>
<tr>
<td>2</td>
<td>10 x 10 cm$^2$</td>
<td>Retracted</td>
</tr>
<tr>
<td>3</td>
<td>20 x 20 cm$^2$</td>
<td>10 cm</td>
</tr>
<tr>
<td>4</td>
<td>20 x 20 cm$^2$</td>
<td>Closed</td>
</tr>
</tbody>
</table>

Neutron production

Through the FLUKA Usrbin scoring card it is possible to investigate the photonuclear reactions on a region by region basis. This method gives an approximation of the fraction of neutrons produced in a certain region compared to the total photoneutron production.

Tracing the origin of neutrons depositing dose in the patient

Although a large fraction of neutrons may be produced in a certain component of the linac, it may be that many of these neutrons are scattered or absorbed before reaching the patient. Therefore information about the neutron production alone may not reveal the neutron dose contribution from each linac component. In a Monte Carlo simulation where neutrons deposit dose in a phantom it is possible to trace back where these neutrons were produced. This has been done with the FLUKA code via application and modification of the following user routines:

1) The mgdraw.f routine is modified to detect photonuclear reactions. Each time such a reaction occurs the location of the reaction is recorded, and each of the reaction products carry this information with them. The Stuprf.f routine ensures that the information is inherited throughout the simulation.

2) The comscw.f routine is called every time energy is deposited in a region of interest. The routine has been modified to record dose originating from neutrons and to extract the information on where these neutrons were created.

The principle of the neutron trace Monte Carlo simulation is illustrated in Figure 2.20. The trace volumes were defined as a 10 x 10 x 10 cm$^3$ cubes placed in depth 0 cm to
2 Experimental set-up

10 cm. The Source to Surface Distance was 90 cm and the beam quality was 15 MV. Trace volumes were defined inside and outside the field to separately investigate the origin of neutrons depositing dose inside and outside the photon field. The center of the out-of-field trace region was at 15 cm lateral distance from the isocenter. For the simulation with 3 x 3 cm\(^2\) photon field, the in-field trace volume was reduced to 3 x 3 x 10 cm\(^2\).

![Diagram of neutron trace volumes](image)

Figure 2.20: This figure is an illustration of the principles behind the neutron origin trace in the Monte Carlo simulations. Neutrons are detected at their origin in the components of the linear accelerator and followed as they pass through the geometry. When neutrons deposit dose in the neutron trace volumes information about their origin is available. The origin of neutrons inside and outside the photon treatment field was investigated separately using two different trace volumes.
2 Experimental set-up
3. Results from Measurements and Monte Carlo simulations

The experimental results are presented in this chapter. This includes the results from the detector characterization measurements and the neutron fluence obtained from measurements and Monte Carlo simulations. For measurements in ion beams, the neutron fluence is reported as neutrons per cm$^2$ per primary ion, while the neutron fluence in photon therapy is normalized per Monitor Unit (MU).

3.1 Measurements

A number of different SRAM chips were tested for the application as neutron detectors. The results for the final SRAM detector found to be best suited are presented in detail here. The term “SRAM detector” refers to the detector presented in section 2.1.1 which was used for the fluence measurements. More results on the other SRAM detector prototypes can be found in (Larsen, 2011, Velure, 2011).

3.1.1 Physikalisch - Technische Bundesanstalt (PTB)

**SRAM detector calibration**

SRAM chips were irradiated with monoenergetic neutron beams of energies 5.8, 8.5 and 14.8 MeV at PTB in Braunschweig. The results from the SRAM detector irradiation were not usable due to a firmware issue at this stage. However, the results from a 1 Mbit SRAM of similar technology (130 nm) revealed the energy dependence of the SEU induction mechanism in this energy range. The SEU cross sections for neutron energies of 5.8 and 8.5 MeV were found to be 14% and 29% of the SEU cross section at 14.8 MeV, respectively.

**BDT Fast neutron calibration**

The detectors were repeatedly irradiated in the three neutron beams of energies 5.8 MeV, 8.5 MeV and 14.8 MeV. The cross section found for the three beam energies are summarized in Table 3.1.
3 Results from Measurements and Monte Carlo simulations

Table 3.1: Results from the fast neutron irradiation of the BDT bubble detectors.

<table>
<thead>
<tr>
<th>Neutron energy [MeV]</th>
<th>5.8</th>
<th>8.5</th>
<th>14.8</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cross section</td>
<td>9.2±2.8</td>
<td>7.8±2.4</td>
<td>7.7±2.9</td>
</tr>
<tr>
<td>[10^{-5} b/(n cm^{-2})]</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

3.1.2 Oslo Cyclotron Laboratory (OCL)

At the Oslo cyclotron laboratory SRAM chips were tested in a 26 MeV proton beam. For the SRAM detector a cross section of (4.5±0.5) \times 10^{-6} SEU/(n cm^{-2}) was found. The irradiations showed that SRAM chips of technologies 90 nm produced high count rates, but possible issues with latch-up errors made measurements difficult to perform and analyze. In the experiment this was observed as sudden abnormal increase in the count rate. SRAMs of this technology were therefore not tested further in this study.

3.1.3 CERN-EU Reference Field (CERF)

The radiation environment at the CERN-EU Reference Field gave the opportunity to look at the response of the SRAM detector to high energy hadrons (HEH) with average energy on the order of 100 MeV (in the assumed cross section plateau region). The cross section for the detector was found to be (9.3 ±1.9) \times 10^{-6} SEU/(HEH cm^{-2}) (Larsen, 2011).

3.1.4 Institute for Energy Technology (IFE), Kjeller

**SRAM detector calibration**

The measurements at Kjeller indicated an upper cross section limit of (8±2) \times 10^{-8} SEU/(n cm^{-2}) for thermal neutrons. This upper limit is calculated assuming that all detected SEUs come from thermal neutron. However, there is reason to believe that higher energy neutrons have contributed to the statistics. This means that the cross section for thermal neutron for this particular detector has been measured to be about
3 Results from Measurements and Monte Carlo simulations

a factor of 100 smaller than for fast neutrons, and may also be significantly smaller than this.

**BDT calibration**

The results from the BDT measurements indicated a cross section for thermal neutrons of \((3.9 \pm 1.0) \times 10^{-5}\) bubbles/(n cm\(^2\)) corresponding to a dose response of \((3.7 \pm 1.0)\) bubbles/\(\mu\)Sv. The reported average sensitivity from the vendor for the detectors was \((2.8 \pm 0.6)\) bubbles/\(\mu\)Sv. These results agree within the uncertainties, and a part of the reason for the higher response in this work may be the exposure of the detector on its way down and up from the measurement position. Future experiments should be performed with longer exposure times at a position with a lower flux, thus reducing these uncertainties.

A summary of the results from the BDT calibration measurements is given in Figure 3.1. The BDT detectors’ average sensitivity to fast neutrons was found to be \(8.2 \times 10^{-5}\) bubbles/(n cm\(^2\)). This is about twice as high as the sensitivity to thermal neutrons in terms of response per fluence. However, the bubble detectors are pre-calibrated from the vendor to measure equivalent neutron doses. Therefore, because the biological effect of neutrons is strongly energy dependent, it may not be the sensitivity in terms of bubbles per neutron which is of most interest, but the response to a deposited equivalent neutron dose. In these terms the measurements at PTB and IFE (Figure 3.1) indicates that the BDT detectors are about a factor of 21 more sensitive to a certain ambient dose equivalent deposited by thermal neutrons compared to that of fast neutrons.

The calibration of the BDT detectors has revealed that the response from fast neutrons must be considered when measuring dose from thermal neutrons. If the fast neutron fluence is known, it may be possible to include corrections for fast neutrons to achieve better accuracy in the calculation of the dose from thermal neutrons.
Results from Measurements and Monte Carlo simulations

Figure 3.1 Measurements: To the left a summary of the cross section measurements for the BDT bubble detector is shown. The results are based on measurements performed at IFE and PTB Braunschweig. The second figure shows the cross section for BDT bubble detectors in units of bubbles per deposited ambient dose equivalent as a function of energy.

TLD calibration

The TLD-600 and TLD-700 were irradiated with thermal neutrons at IFE. The TLDs were found to have a thermal neutron cross section of $(5.9 \pm 1.4) \times 10^{-6} \text{nC/(n cm}^2\text{)}$, or in terms of dose; $557 \pm 128 \text{nC/mSv}$. The data analysis revealed that even with about 5 cm of lead shielding, 80% of the response in the TLD-600 originate from photons. This indicates that a thicker lead shielding may increase the accuracy of the calibration procedure.

3.1.5 The Svedberg Laboratory (TSL), Uppsala

SRAM detector calibration

The SRAM detector was exposed to a 174.5 MeV quasi-monoenergetic neutron (QMN) beam. Four identical SRAM detectors (each with four SRAM chips) were tested. The average SEU cross section was found to be $(9.9 \pm 1.1) \times 10^{-6} \text{SEU/(n cm}^2\text{)}$.

In Figure 3.2 the measured neutron cross section for each chip can be seen. From this data a mean response of $2.5 \times 10^{-6} \text{SEU/(n cm}^2\text{)}$ and a standard deviation of 11% are found. The results for each detector are given in Table 3.2.
3 Results from Measurements and Monte Carlo simulations

Figure 3.2 - Measurements: SEU cross section of the individual SRAM chips in the neutron beam. The dashed line shows the mean cross section of $2.5 \times 10^{-6}$ SEU/(n cm$^{-2}$).

Table 3.2 - Measurements: The table shows the found neutron cross sections for the four SRAM detectors which were tested in the neutron beam.

<table>
<thead>
<tr>
<th>Detector</th>
<th>Cross section [SEU/(n cm$^{-2}$)]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>$(9.4 \pm 1.0) \times 10^{-6}$</td>
</tr>
<tr>
<td>2</td>
<td>$(1.0 \pm 0.1) \times 10^{-5}$</td>
</tr>
<tr>
<td>3</td>
<td>$(1.1 \pm 0.1) \times 10^{-5}$</td>
</tr>
<tr>
<td>4</td>
<td>$(9.2 \pm 1.0) \times 10^{-6}$</td>
</tr>
</tbody>
</table>

The impact of the angle of the incoming beam was also investigated. The results from these measurements are presented in Table 3.3 and illustrate that the cross section remains similar for different irradiation angles.

Table 3.3 - Measurements: The table shows the average response to neutrons for the SRAM detectors for different irradiation angles.

<table>
<thead>
<tr>
<th>Irradiation angle</th>
<th>Sensitivity [SEU/(n cm$^{-2}$)]</th>
</tr>
</thead>
<tbody>
<tr>
<td>$0^\circ$</td>
<td>$(9.9 \pm 1.1) \times 10^{-6}$</td>
</tr>
<tr>
<td>$45^\circ$</td>
<td>$(9.7 \pm 1.2) \times 10^{-5}$</td>
</tr>
<tr>
<td>$180^\circ$</td>
<td>$(9.0 \pm 0.8) \times 10^{-6}$</td>
</tr>
</tbody>
</table>
Summary of the SRAM detector calibration

The energy dependence of the SEU mechanism has been investigated as a part of this project in cooperation with Eivind Larsen and Arild Velure. The results from the calibration campaign are summarized in Figure 3.3 and in Table A-2 in the appendix.

![SRAM detector cross section summary](image)

*Figure 3.3 - Measurements: This plot summarizes the SRAM detector characterization measurements. The neutron cross section is plotted as a function of neutron energy. The data point at 26 MeV is for protons and the data point at 100 MeV is for a mixed field of particles with a wide energy range as indicated by the horizontal error bars.*

It can be seen that the results from CERF and TSL agree well. This may indicate that the response to charged particles and neutrons in the energy range of these two experiments may be similar. The results also demonstrate that the SEU cross section decreases at lower energies as expected. However, uncertainties are present, especially at low energies. The data points at 5.8 MeV, 8.5 MeV and 14.8 MeV are based on measurements at PTB with a different SRAM detector. By assuming that the SEU mechanisms in both detectors have the same energy dependence, the neutron cross section for the main SRAM detector has been estimated for the neutron energies applied at PTB. Further, the data point at 26 MeV is for protons, and the neutron cross section at this energy may be higher due to the impact of the Coulomb barrier at
this energy. More details on the calibration of the SRAM detector can be found in (Larsen, 2011, Velure, 2011).

The SRAM detector calibration campaign indicates that the detector can be well suited for applications in particle therapy. Especially considering that average neutron energies may about half the primary beam energy (Gunzert-Marx et al., 2008). Furthermore, when in-phantom neutron measurements are desired, the SRAM detector may be an alternative to passive detectors. With more knowledge on the cross section for charged particles, the contribution from these in the measurement positions may be better accounted for either through measurements of the charged particle flux or dedicated Monte Carlo simulations.

**Proton beam - Neutron fluence measurements**

In Figure 3.4 the measured neutron fluence at Bragg peak depth in the TSL proton beam experiment is shown. The fluence is plotted as a function of lateral distance from the beam axis. We see that the measured neutron fluence drops rapidly with increasing lateral distance from the beam axis. For comparison FLUKA Monte Carlo simulations of the total neutron fluence as well as the neutron fluence above 1 MeV is shown. The fluence is given as neutrons per cm\(^2\) per primary proton, and the measured fluence ranged from $8.6 \times 10^{-3}$ n cm\(^{-2}\) pr\(^{-1}\) to $4.9 \times 10^{-5}$ n cm\(^{-2}\) pr\(^{-1}\). However, comparing the measurements with the simulations it is reasonable to assume that the measurement closest to the beam axis has been significantly affected by charged particles. In general, the SRAM detector reports neutron fluences between the two Monte Carlo simulation curves. Because the SRAM detection threshold is on the order of 1 MeV the measured fluence was expected to be similar to the fluence obtained from the Monte Carlo simulation with this threshold. The higher fluence obtained from the measurements may indicate contributions to the detector signal from charged particles. The measurement in the position at 0.8 cm lateral distance from the beam axis (described in section 2.2.5) could not be conducted because the beam intensity was too high.
Results from Measurements and Monte Carlo simulations

Figure 3.4 – Measurements and Monte Carlo simulations: The figure shows the detected neutron fluence per primary proton as a function of lateral distance from the beam axis. The Monte Carlo simulations show one curve including all neutrons and one curve for neutrons above 1 MeV only.

The neutron cross sections found in the expected plateau region of the SRAM detector (Table 3.2) have been used as a basis for the fluence calculations. This introduces an error in the fluence calculation, but is considered the best approach with the information currently available concerning the SRAM detector’s cross section for neutrons. The error in the fluence from this assumption was estimated to be on the order of 30% for the measurement position at 5.2 cm lateral distance from the beam axis.

Neutron fluence measurements were also performed at six different depths prior to and deeper than the Bragg peak. These measurements were conducted for lateral distances of 5.2 cm, 9.5 cm and 13.7 cm. The results from the measurements are presented in Figure 3.5, where a separate curve is drawn for each of the measurement series. The measurements show a relatively small change in neutron fluence with depth. The plot also confirms the findings visible in Figure 3.4 which show that there is a rapid decrease in neutron fluence for increasing lateral distance to the beam axis. Monte Carlo simulations are also shown, illustrated by the dotted and dashed lines. The simulation results are shown separately for neutrons with energies above 1 MeV.
3 Results from Measurements and Monte Carlo simulations

Figure 3.5 – Measurements and Monte Carlo simulations: Neutron fluence as a function of depth measured with the SRAM detector at three different lateral distances. The solid lines represent the measurements and indicate little reduction of neutron fluence with depth. For comparison, Monte Carlo simulations of total neutron fluence (dotted lines) and for neutrons above 1 MeV (dashed lines) are shown. The fluence is reported per primary proton.

Overall, the Monte Carlo simulations and the measurements in Figure 3.5 show the same behavior, but the results show that the SRAM detector reports fluence just below the total neutron fluence estimated by the Monte Carlo simulations. Considering the uncertainties in the measurements represented by the error bars, there is reasonable agreement between the measurements and the Monte Carlo simulations.

For the measurements closest to the beam axis, illustrated by the black line with circular markers, the results indicate that the detector may have received a significant contribution from charged particles for depths more shallow than the Bragg peak depth. The detected fluence here is higher than expected from the Monte Carlo simulation. This effect is not to the same degree visible at larger lateral distances, and this is in agreement with the assumption that the relative charged particle contribution is decreasing with lateral distance.

The results from the angular fluence measurement are shown in Figure 3.6. The neutron fluence ranges from $4 \times 10^{-5}$ n cm$^{-2}$ per proton at $0^\circ$ to $10^{-5}$ n cm$^{-2}$ per proton.
3 Results from Measurements and Monte Carlo simulations

at 90°. These values are comparable to fluences measured at 13.7 cm distance from the beam axis inside the phantom at various depths. Although the measurements at 30° and 45° deviate from the general trend of the measurements, reasonable agreement can be observed between the Monte Carlo simulations and the measurements considering the uncertainties involved. The observed change in fluence with angle is relatively small compared to the steeper fluence gradients measured inside the phantom. Repeated measurements may reveal if the observed deviations are caused by set-up errors or other sources of error.

Figure 3.6 – Measurements and Monte Carlo simulations: Neutron fluence per primary proton as a function of angle relative to the beam axis. The measurements were performed 20 cm behind the Bragg peak. The results indicate that the neutron fluence is reduced by a factor of four from 0° to 90°.

3.1.6 GSI Helmholtz Centre for Heavy Ion Research

**SRAM neutron fluence measurements**

Measurements of neutron fluence with the SRAM detector were performed with a water phantom irradiated by a 200 MeV/u carbon beam at GSI. For the majority of the allotted time for experiments, the primary beam was unstable, this in turn resulting in unstable and to some degree unreliable results. The suspicions of an unfocused beam were confirmed through irradiation of a Gaf-Chromic film (Figure
3 Results from Measurements and Monte Carlo simulations

3.7). Although this makes it difficult to conclude regarding the neutron fluence, the measurements may still give useful information on the feasibility of neutron fluence measurement with the SRAM detector.

Results from the angular neutron fluence measurements are presented in Figure 3.8. Because there is a significant fraction of charged fragments in these positions, corrections based on the charged particle fluence from FLUKA Monte Carlo simulations have been applied. It is assumed in the corrections that the charged particle SEU cross section is the same as for neutrons.

![Image](image.png)

*Figure 3.7: Irradiation of a Gaf-Chromic film revealed that the primary beam was unfocused and shifted towards one side at the end of the experiment. One square is 5 x 5 mm².*

In Figure 3.9 neutron fluence as a function of lateral distance from the beam is shown. The detectors were placed on opposite sides of the water phantom. The difference in response for the two detectors indicates the impact of the unstable beam. Irradiation of the Gaf-Chromic film confirmed that the primary beam, at the time of the measurements with the Gaf-Chromic film, was unfocused and smeared out towards the position of SRAM detector 1 in Figure 3.9. This may add to the explanation of the high response in this detector.

Due to the unstable beam during the experiment it was not possible to conclude concerning the neutron fluence from the measurements. Still, the measurements demonstrate that the SRAM detector may be a candidate for neutron dosimetry in such an environment.

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3 Results from Measurements and Monte Carlo simulations

Figure 3.8 – Measurements and Monte Carlo simulations: Angular neutron fluence distribution from measurements and FLUKA Monte Carlo simulations.

**Measurements with an antropomorphic phantom**

For the delivered dose of 2 Gy (8.64 x 10^8 carbon ions) to the target volume, 92 SEUs were detected. This gives a first estimate of the neutron fluence of 9.3 x 10^6 n cm^{-2}. However, FLUKA Monte Carlo simulations indicated that only 35% of the hadrons with energies above 1 MeV were neutrons. Correcting the neutron fluence, assuming that the SEU cross section is the same for all particle species, gives a neutron fluence of 3.3 x 10^6 n cm^{-2}. Considering the statistical error and uncertainties in the SEU cross section this is equivalent to a neutron fluence of (1.6±0.4) x 10^6 n cm^{-2} Gy^{-1} or (3.7±0.9) x 10^{-3} n cm^{-2} per primary carbon ion. FLUKA Monte Carlo simulations of neutrons above 1 MeV gave a fluence of 3.96 x 10^{-3} n cm^{-2} pr^{-1}. The results indicate that the SRAM detector may be useful for measurements in such experiments provided that information on the charged particle fluence is available. The signal from the scintillation detectors in the set-up provided a measure of the average charged particle fluence over the area covered by the scintillators. However, due to the inhomogeneous radiation field with sharp dose gradients in the measurement area it was clear that using the average fluence over the scintillator area did not provide the accuracy needed to provide data for the subtraction of the SRAM detectors.
response to charged particles. A dedicated charged particle detector with comparable size to the SRAM chips may however give useful information and improve the accuracy of neutron fluence measurements with the SRAM detector.

![Figure 3.9](image)

*Figure 3.9 – Measurements and Monte Carlo simulations: The figure shows the neutron fluence as a function of lateral distance from the beam axis. The two SRAM detectors were placed one opposite sides of the beam axis. The large difference in detectors response indicates that the primary carbon beam was positioned closer to SRAM detector 1. The horizontal error bars are based on results from irradiation of the Gaf-Chromic film and give only a rough estimate of effect from the defocused and misaligned beam.*

### 3.1.7 Haukeland University Hospital

**Neutron fluence and energy spectra measurements with BDS bubble detectors**

Measurements were performed in 11 different positions inside and outside the water phantom. The neutron energy spectra were all unfolded using the BUNTO code (Ongaro et al., 2001). In Figure 3.10 obtained spectra for depths between 0 cm and 10 cm are shown. These spectra are measured at 4 cm lateral distance to the photon field edge. The spectra are relatively similar in their energy distribution while the integrated fluence varies with more than one order of magnitude.
Results from Measurements and Monte Carlo simulations

At approximately 1 MeV a peak is visible in all the spectra. The peak corresponds well to the expected evaporation peak, while the contribution from direct knock-out neutrons, expected at higher energies, is relatively small in comparison. There is a clear dip in the spectra in the energy interval covered by the BDS 100 detector (100 keV – 600 keV). Comparing to neutron spectra obtained in the literature under similar conditions (Howell et al., 2009, Kaderka, 2011) it may be suspected that the fluence in this detector region has been underestimated. This may again have led to an overestimation of the neutron fluence in the lower energy bins. Additional measured energy spectra are shown in Figure A-1 in the appendix.

![Neutron energy spectra at different depths](image)

*Figure 3.10 - Measurements: Neutron energy spectra for 5 different depths in water ranging from surface measurements to measurements at 10 cm depth. All spectra are measured at 4 cm lateral distance from the photon field edge. In the last plot all positions are shown illustrating how the fluence decreases with increasing depth.*

The measured neutron fluences for all positions are summarized in Table 3.4. The results are normalized to one Monitor Unit (MU). At the surface the fluence was
3 Results from Measurements and Monte Carlo simulations

observed to be \((2.2\pm0.5) \times 10^5\) n cm\(^{-2}\) MU\(^{-1}\), and the fluence decreases to \((6.0\pm1.2) \times 10^3\) n cm\(^{-2}\) MU\(^{-1}\) at 10 cm depth. The results are plotted as functions of depth and lateral distance from the field edge in Figure 3.11. We see that the fluence does not change significantly with lateral distance from the field edge, but drops rapidly with increasing depth. This indicates that the linac head is the primary source of neutrons.

Table 3.4 – Measurements: The neutron fluence per Monitor Unit (MU) obtained with the BDS spectrometer.

<table>
<thead>
<tr>
<th>Depth in Phantom [cm]</th>
<th>Lateral distance from field edge [cm]</th>
<th>Neutron fluence [n cm(^{-2})MU(^{-1})]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td></td>
<td>((2.2 \pm 0.5) \times 10^5)</td>
</tr>
<tr>
<td>3</td>
<td></td>
<td>((9.7 \pm 1.9) \times 10^4)</td>
</tr>
<tr>
<td>5</td>
<td>4</td>
<td>((4.0 \pm 0.8) \times 10^4)</td>
</tr>
<tr>
<td>8.6</td>
<td>4</td>
<td>((1.0 \pm 0.2) \times 10^4)</td>
</tr>
<tr>
<td>10</td>
<td></td>
<td>((6.0 \pm 1.2) \times 10^3)</td>
</tr>
<tr>
<td>3</td>
<td>8</td>
<td>((8.0 \pm 1.6) \times 10^4)</td>
</tr>
<tr>
<td>5</td>
<td>8</td>
<td>((3.2 \pm 0.6) \times 10^4)</td>
</tr>
<tr>
<td>8.6</td>
<td>8</td>
<td>((9.4 \pm 1.9) \times 10^3)</td>
</tr>
<tr>
<td>3</td>
<td>12</td>
<td>((8.1 \pm 1.6) \times 10^4)</td>
</tr>
<tr>
<td>5</td>
<td>12</td>
<td>((3.6 \pm 0.7) \times 10^4)</td>
</tr>
<tr>
<td>8.6</td>
<td>12</td>
<td>((1.0 \pm 0.2) \times 10^4)</td>
</tr>
</tbody>
</table>

Figure 3.11 - Measurements: Left plot: Neutron fluence as a function of depth in water. The plot indicates that the neutron fluence decreases exponentially with depth in water. Right plot: Fast neutron fluence as a function of lateral distance from the field edge for three different depths in water. The results suggest that there is little correlation between the neutron fluence and the lateral position within the measurement range.
With the application of the BUNTO code, the BDS detectors cover the energy range from 10 keV and up to 25 MeV. This means that neutrons moderated to energies below 10 keV are no longer visible for the spectrometer. This effect may cause an underestimation of the fluence increasing with depth as a larger fraction of the neutrons are moderated below the detector threshold. However, information on the lower energy neutrons is available through measurements with the BDT (bubble detector thermal) and TLDs as well as in the Monte Carlo simulations performed in this work.

**Thermal neutron measurements with BDT and TLD detectors**

The thermal neutron fluences obtained from measurements with Thermoluminescence (TLD) detectors and the thermal bubble detectors (BDT) are presented in Figure 3.12 and Figure 3.13. Measurements with the TLDs indicate that the thermal neutron fluence increases for the first few cm in the phantom and then is reduced with increasing depth. This may be expected as thermal neutrons are produced through moderation of higher energy neutrons as they interact inside the phantom. The estimated thermal neutron fluence from the TLD measurements ranged from \((1.6 \pm 0.4) \times 10^5\) n cm\(^{-2}\) MU\(^{-1}\) to \((4 \pm 1) \times 10^4\) n cm\(^{-2}\) MU\(^{-1}\). The lowest thermal neutron fluence was found at the surface of the phantom, while the highest value was found at 3 cm depth and 4 cm lateral distance from the field edge.

The analysis of the BDT measurements (Figure 3.12) yields significantly higher thermal neutron fluences at shallow depths compared to the TLD measurements. The investigation of the BDT characteristics showed that the BDT detectors will be affected by fast neutrons. Based upon prior knowledge about the BDT’s sensitivity to fast neutrons, as measured at PTB, it is reasonable to believe that a significant fraction of the signal in the BDT detectors is not due to thermal neutrons, but fast neutrons. As Figure 3.11 (a) indicates, the fast neutron fluence drops rapidly with depth, and the error in the BDT measurements is therefore expected to be more pronounced for shallow depths. This is also where we see clear disagreement between
the TLD and BDT results. The BDT and TLD measurements agree within the uncertainties for positions at 5 cm depth and deeper.

A correction could perhaps be applied to the BDT measurements. However, it is difficult to apply an accurate correction to the data without knowledge about the BDTs sensitivity to the neutrons in the energy range between thermal and the calibration energies at PTB (5.8 MeV and higher energies). Figure 3.13 shows the thermal neutron fluence as a function of lateral distance from the field edge. Although a slight drop in the fluence is observed with increasing lateral distance, the results indicate that the thermal neutron fluence changes only to a small degree in the lateral direction. These results are similar to what was found for fast neutrons with the BDS detectors. The modest fluence reduction at large lateral distances could also be due to the fact that these positions are close to the phantom edge and therefore it may be that fewer neutrons scatter from the surrounding medium and into the measurement positions.
3 Results from Measurements and Monte Carlo simulations

Figure 3.13: Thermal neutron fluence as a function of lateral distance from the field edge for three different depths in water. Results from the TLDs are shown in the left plot, while the plot to the right shows the BDT measurements.

3.2 Monte Carlo Simulations

3.2.1 Proton beams

Neutron energy spectra for three of the measurement positions in the proton beam experiments are shown in Figure 3.14. The spectra are plotted in isolethargic units¹ and the area beneath the curve in any energy interval is therefore proportional to the neutron fluence in the interval. Thus we can see that neutrons with energies above 10 MeV dominate the spectra close to the beam axis, while the fraction of thermal neutrons increases for larger lateral distances inside the phantom.

A Monte Carlo study with a 107 MeV proton beam was performed in order to investigate the neutron fluence at lower treatment energies. The results are compared to the results for neutron fluence from 178 MeV proton beam Monte Carlo simulations in Figure 3.15 and Figure 3.16. The results from the Monte Carlo simulations show that the neutron fluence is reduced when the proton energy is

¹ The use of isolethargic plots includes a change in variable by instead of plotting a function $F(x)$ as a function of $x$, $(dF(x)/dx)$, $F(x)$ is plotted as a function of the logarithm of $x$, $dF(x)/d\log(x)$. Presenting the data with logarithmic x-axis and linear y-axis ensures then that the area beneath the curve between any pair of abscissas $\log(x_1)$ and $\log(x_2)$ is proportional to the integral of $F(x)$ between the values $x_1$ and $x_2$. 

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3 Results from Measurements and Monte Carlo simulations

lowered. Figure 3.15 indicates that the relative difference in neutron fluence increases with lateral distance from the beam axis. At the beam axis and at the Bragg peak depth, the fluence is about 19% higher for the 178 MeV proton beam, while at 13.7 cm lateral distance the fluence is more than 3 times higher for the 178 MeV proton beam compared to the 107 MeV proton beam. In Figure 3.16 the same trend can be seen. From this figure, we can also observe that the neutron fluence decreases relatively more with depth beyond the Bragg peak for the lower primary beam energy.

Figure 3.14 – Monte Carlo simulations: Neutron energy spectra at Bragg peak depth for three different lateral distances from the beam axis. We see that neutrons of energies above 10 MeV dominate the spectra. The spectra were obtained from Monte Carlo simulations of the 178 MeV proton beam experiment.
3 Results from Measurements and Monte Carlo simulations

Figure 3.15 – Monte Carlo simulations: The figure shows the simulated total neutron fluence as a function of lateral distance from the beam axis at Bragg peak depth for the proton beam energies 107 MeV and 178 MeV. The relative difference between the two curves increases with lateral distance from the beam axis.

Figure 3.16 – Monte Carlo simulations: The figure shows the simulated total neutron fluence as a function of depth in water relative to the Bragg peak for the proton beam energies 107 MeV (dotted lines) and 178 MeV (solid lines). The neutron fluence is on average about 2.5 times higher for the 178 MeV proton beam.

Two dimensional plots of neutron and proton fluence from the Monte Carlo simulations are presented in Figure 3.17 and Figure 3.18. The latter plot illustrates that protons are present also far from the beam axis, and we see that the proton fluence quickly is reduced to more than one order of magnitude less than the neutron...
3 Results from Measurements and Monte Carlo simulations

fluence. The neutron fluence drops by approximately two orders of magnitude from the Bragg peak area at the beam axis to 15 cm lateral distance to the Bragg peak, while the proton fluence is reduced by a factor of $10^6$ over the same distance. Figure 3.19 shows more quantitatively how the neutron fluence is higher than the proton fluence for lateral distances of more than approximately 3 cm from the beam axis.

Figure 3.17 - Monte Carlo simulations: Neutron fluence per primary proton in the water phantom. The neutron fluence ranges three orders of magnitude from the primary beam axis to the edge of the phantom.

Figure 3.18 - Monte Carlo simulations: Proton fluence per primary proton in the phantom. We see that the proton fluence decreases very fast at the Bragg peak and is reduced by more than six orders of magnitude for large lateral distances.
3 Results from Measurements and Monte Carlo simulations

![Particle fluence at Bragg peak depth - 178 MeV protons](image)

Figure 3.19 - Monte Carlo simulations: Proton and neutron fluence as a function of lateral distance from the beam axis. The primary proton beam energy was 178 MeV, and the plot shows that the neutron fluence is not reduced as rapidly with increasing distance from the beam axis as the proton fluence.

3.2.2 Carbon ion beams

Neutron energy spectra for three different lateral distances from the beam axis are shown in Figure 3.20. We can see that neutrons with energies above 10 MeV dominate the spectra, while the fraction of thermal neutrons increases for larger lateral distances inside the phantom as observed in the spectra from the proton beam. However, it is clear that the neutrons from the carbon ion beam extend to higher energies.

In Figure 3.21 the neutron fluence from 341 MeV/u and 200 MeV/u carbon ions at the Bragg peak depth is plotted as a function of lateral distance from the beam axis. At the beam axis the neutron fluence is 21% higher for the 341 MeV/u beam compared to the 200 MeV/u beam. The neutron fluence is also reduced faster for the lower energy when moving away from the beam axis. In Figure 3.22 the neutron fluence is shown as a function of depth for the two energies. Again we see that the neutron fluence is reduced with the primary energy of the carbon ions. On average the relative neutron fluence was found to be approximately a factor of 3 lower for the
3 Results from Measurements and Monte Carlo simulations

200 MeV/u carbon ions in the measurement positions. Because the difference is less pronounced in the areas with high fluence, the absolute difference in neutron yield from the two beam energies is significantly less than this.

![Image of neutron energy spectra](image1)

**Figure 3.20 - Monte Carlo simulations: Neutron energy spectra at Bragg peak depth for three different lateral distances from the beam axis. The spectra were obtained from Monte Carlo simulations of a 341 MeV/u carbon ion beam.**

![Image of neutron fluence](image2)

**Figure 3.21 - Monte Carlo simulations: The figure shows the simulated total neutron fluence as a function of lateral distance from the beam axis at Bragg peak depth for the carbon ion beam energies 200 MeV/u and 341 MeV/u. At the beam axis the fluence is about 21% higher for the 341 MeV/u.**
3 Results from Measurements and Monte Carlo simulations

Figure 3.22 - Monte Carlo simulations: The figure shows the simulated total neutron fluence as a function of depth in water relative to the Bragg peak for the carbon beam energies 200 MeV/u (dotted lines) and 341 MeV/u (solid lines). The neutron fluence is on average approximately 3 times higher for the 341 MeV/u beam.

A two dimensional plot of the neutron fluence from the Monte Carlo simulation of 341 MeV/u carbon ions is presented in Figure 3.23. We see that the neutron fluence ranges over three orders of magnitude within the phantom, with its maximum at the primary beam axis at a depth prior to the Bragg peak.

Figure 3.23 - Monte Carlo simulations: Neutron fluence per primary carbon ion in the phantom. The neutron fluence ranges three orders of magnitude from the primary beam axis to the edge of the phantom.
3 Results from Measurements and Monte Carlo simulations

3.2.3 Photon beams

**Verification of the Monte Carlo linac model**

Monte Carlo Simulations and measurements of the photon depth dose along the central axis from a 10 x 10 cm$^2$ photon field are shown in Figure 3.24. The Monte Carlo simulations show good agreement with the measured data.

![Photon depth doses](image1)

*Figure 3.24: Depth dose curves for a 10x10 cm$^2$ field with a 15 MV photon beam from measurements with an ionization chamber and from FLUKA Monte Carlo simulations. A good agreement between measurements and simulations is observed.*

![Cross plane photon dose profiles and In Plane photon dose profiles](image2)

*Figure 3.25: Cross plane (left) and in plane (right) dose profiles for a 10x10 cm$^2$ field with a 15 MV photon beam from measurements and Monte Carlo simulations.*
3 Results from Measurements and Monte Carlo simulations

In addition, Monte Carlo simulations and measurements of the in plane and cross plane beam profiles were conducted (Figure 3.25). The dose profiles show that the simulations match the measurements well, especially when considering that the scope of this Monte Carlo model is to estimate neutron production and not clinical photon doses.

Monte Carlo simulations also yielded that $5.45 \times 10^{14}$ electrons are needed to deliver 1 Gy to the isocenter. With the settings used in the Monte Carlo simulations, 130 MU are needed to deliver 1 Gy to the isocenter for the linac used in this work. This means that $4.19 \times 10^{12}$ primary electrons are needed to deliver 1 MU.

**The origin of neutron production**

The results from Monte Carlo simulations of neutron production in the linac are presented in Figure 3.26. The results indicate that about 50% of the neutrons are produced in the primary collimator. The Y-jaws also produce significantly more neutrons than the X-jaw as may be expected, considering that the Y-jaws are placed above the X-jaws. The MLCs have very little influence on the neutron production.

![Regions of neutron production](image)

*Figure 3.26 – Monte Carlo simulations: Relative neutron production for the linac components with four different collimator settings. This plot illustrates in which components (regions) of the linac neutrons are produced.*
while retracted, but when the MLCs are applied to shape the field, the relative neutron production in the MLCs is observed to increase, as expected.

In Table 3.5 the results from this work is compared to the results from a MCNPX Monte Carlo simulation of a Varian 2100C linac in the same energy mode (15 MV). The results are in good agreement indicating that the primary collimator is the largest contributor to neutron production.

Table 3.5: An overview of the neutron production region by region for 10 x 10 cm² photon field size. Data for comparison is taken from (Ma et al., 2008) simulating the neutron yield from a Varian Clinac 2100C in the 15 MV mode using the MCNPX Monte Carlo code.

<table>
<thead>
<tr>
<th>Position</th>
<th>This work</th>
<th>(Ma et al., 2008)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Primary collimator</td>
<td>49.2%</td>
<td>57.9%</td>
</tr>
<tr>
<td>Target (tungsten)</td>
<td>4.1%</td>
<td>5.0%</td>
</tr>
<tr>
<td>Target (copper)</td>
<td>6.6%</td>
<td>5.6%</td>
</tr>
<tr>
<td>Flattening filter</td>
<td>1.9%</td>
<td>1.4%</td>
</tr>
<tr>
<td>Y-Jaws</td>
<td>25.1%</td>
<td>22.6%</td>
</tr>
<tr>
<td>X-Jaws</td>
<td>12.5%</td>
<td>7.4%</td>
</tr>
<tr>
<td>MLCs</td>
<td>0.02%</td>
<td>0.03%</td>
</tr>
<tr>
<td>Remaining components</td>
<td>0.6%</td>
<td>-</td>
</tr>
</tbody>
</table>

A summary of the results from the neutron trace Monte Carlo simulations is presented in Figure 3.27. The average relative neutron contribution for each linac component for the 4 collimator settings is shown. The relative neutron contribution is reported in terms of neutron production, neutron dose deposition in the treatment field and neutron dose deposition outside the treatment field. It can be seen that the relative neutron production is higher than the relative neutron dose contribution for the components far from the water phantom, such as the primary collimator and the target. For the Y-jaws the relative neutron production alone seems to give a good estimate for the neutron dose contribution in this particular case. For the components closer to the phantom (X-jaws and MLCs) the relative neutron dose contribution in the phantom is larger than expected from considering only the neutron production in these components.
3 Results from Measurements and Monte Carlo simulations

![Graph showing average regional neutron contributions for different components.](image)

**Figure 3.27 – Monte Carlo simulations**: The plot shows the average results for the four collimator settings used in the Monte Carlo simulations. The three bars represent the neutron production and the neutron dose contribution for the different linac components. While the neutron production is highest in the primary collimator, neutrons from other components such as the jaws and MLCs may contribute more to the neutron dose in the patient.

The analysis of the results indicates that the neutron production in a component and the neutron dose contribution from the component are not directly comparable. Further, there is also a difference in the neutron dose contributions inside and outside the treatment field. The primary collimator, the target and the flattening filter show a higher in-field dose contribution compared to the out-of-field dose contribution. On the other hand, the jaw-collimators and the MLCs seem to be especially important for out-of-field neutron doses.

Studying the results for the different collimator settings in Figure 3.28, it is clear that the configuration has a great impact on the dose contribution from the adjustable collimators, the jaws and MLCs. The plots illustrate how the MLCs contribution to neutron dose both inside and outside the treatment field increases as they are applied for beam collimation. Further the results suggest that the components close to the patient may have a major impact on the neutron dose to the patient.
3 Results from Measurements and Monte Carlo simulations

Figure 3.28 – Monte Carlo simulations: Results from the neutron trace Monte Carlo simulations comparing the 4 different collimator settings. The plot to the left illustrates which components contribute the most neutron dose inside the treatment field, while the right plot shows the relative out-of-field neutron dose contribution for each component. We can see that the results are strongly dependent on the collimator settings.

**Neutron fluence results**

In Figure 3.29 the results from Monte Carlo simulations of the neutron fluence in the measurement positions are shown. The solid lines represent the total neutron fluence while the dotted lines show fluence of neutrons above 10 keV (The BDS detection threshold).

Figure 3.29 – Monte Carlo simulations: Neutron fluence in the measurement positions as a function of depth (left) and as a function of lateral distance from the field edge (right). The solid lines represent the total neutron fluence while the dotted lines show the obtained neutron fluence for energies above 10 keV (the BDS detection threshold).
Results from Measurements and Monte Carlo simulations

The simulated fluence in the measurement positions range from $5.9 \times 10^4$ n cm$^{-2}$ MU$^{-1}$ to $2.1 \times 10^4$ n cm$^{-2}$ MU$^{-1}$, where one Monitor Unit is equivalent to $4.19 \times 10^{12}$ primary electrons in the Monte Carlo simulations. The Monte Carlo simulations show the same trend as the measurements regarding the distribution of neutrons in depth and with lateral distance from the treatment field (see Figure 3.11). However, when comparing the absolute values to the experimental results, it is clear that the Monte Carlo simulations predict significantly lower neutron fluences than the measurements. This is also illustrated in Figure 3.30 where the neutron fluence results from measurements and Monte Carlo simulations as a function of depth are shown. The green dashed line represents the values obtained from Monte Carlo simulations while the solid lines represent measurements. On average for all measurement positions, the neutron fluence in the simulations was about a factor of four lower than the measured fluence. The analysis of the measured data and the Monte Carlo simulation results has not yielded any clear reason for this discrepancy.

![Graph showing neutron fluence results](image)

**Figure 3.30 Measurements and Monte Carlo simulations**: Thermal, fast and total neutron fluence obtained with BDS and TLD detectors at 4 cm distance from the field edge. For comparison, FLUKA Monte Carlo simulations of the total fluence in the measurement positions are also shown. The term “fast neutrons” refers to neutrons with energies above 10 keV (The BDS detector threshold).
3 Results from Measurements and Monte Carlo simulations

Figure 3.31 shows a two dimensional plot of neutron fluence inside the phantom. The distance in depth from the isocenter is displayed along the x-axis while the y-axis displays the distance from the beam axis. The colors represent the neutron fluence as described by the color bar on the right side of the plot. The figure illustrates how the neutrons from the treatment head components enter the phantom and how the fluence is reduced with depth.

![Neutron fluence - 15 MV photons](image)

**Figure 3.31 – Monte Carlo simulations: Total Neutron fluence per Monitor Unit from FLUKA Monte Carlo simulation of the Varian 23 iX linac in the 15 MV energy mode. The phantom walls are indicated with the black lines and the applied 3 x 3 cm² photon field enters the phantom from the left side.**

Figure 3.32 shows the neutron energy spectra for three of the experimental measurement positions. The spectra are plotted in isolethargic units. We see that the fast neutron component consists of a broad peak in the energy region from above 10 keV to a few MeV and that this peak is significantly reduced with depth. The thermal neutron fluence is low at the surface and it reaches its maximum at intermediate depths, as observed in the measurements. The spectra agree reasonably with spectra found in the literature both from measurements (Howell et al., 2009) and Monte Carlo simulations (Martinez-Ovalle et al., 2011). Comparing the spectra to the measured spectra (see Figure 3.10) supports the theory that the bubble detectors may be overestimating the fluence at energies below 100 keV.
3 Results from Measurements and Monte Carlo simulations

Figure 3.32 – Monte Carlo simulations: Neutron energy spectra for three different depths in water obtained from Monte Carlo simulations of a 3 x 3 cm$^2$ photon field in the 15 MV energy mode. The spectra are dominated by a fast component in the energy range around 1 MeV and a thermal neutron peak which becomes more pronounced with depth. The error bars show the statistical error in the simulations.

3.3 Summary of out-of-field neutron fluence measurements

3.3.1 Proton beams

The measurements indicate that the neutron fluence is rapidly reduced with increasing distance to the beam axis. Little change in fluence with depth was observed. The detected neutron fluence was on the order of one neutron cm$^{-2}$ per 1000 primary protons or less depending on the position. Neutron spectra are dominated by fast neutrons of energies above 10 MeV. The performance of the SRAM detector in the experiment was promising and the yielded results showed reasonable agreement with the Monte Carlo simulations considering the uncertainties. The results indicate that SRAM detector may be used for neutron fluence measurements in proton therapy.
3 Results from Measurements and Monte Carlo simulations

3.3.2 Carbon ion beams

The measurements with the SRAM detector in carbon ion beams show that the detector may be a useful tool in the assessment of the neutron fluence in carbon ion therapy. FLUKA Monte Carlo simulations indicate that the distribution of neutrons from carbon ions is similar to that from protons. The estimated neutron fluence was on the order of one neutron cm$^{-2}$ per 10 primary ions or less depending on the position. More experimental data would strengthen the findings from the simulations concerning the magnitude of the neutron fluence.

3.3.3 Photon beams

The measurements indicate that the neutron fluence is strongly dependent on depth, while the distance from the treatment field is of less importance. This is in agreement with the assumption that the neutrons originate from the treatment head and not the patient. Adding together the results from the BDS and TLD measurements gives the total measured neutron of $2.6 \times 10^5$ n cm$^{-2}$ MU$^{-1}$ at the surface and $6.4 \times 10^4$ n cm$^{-2}$ MU$^{-1}$ at 10 cm depth. The Monte Carlo simulations indicate significantly lower neutron fluence ranging from $5.9 \times 10^4$ n cm$^{-2}$ MU$^{-1}$ to $2.1 \times 10^4$ n cm$^{-2}$ MU$^{-1}$. Still, the shape of the neutron depth dose curves from measurements and simulations show the same trend.
3 Results from Measurements and Monte Carlo simulations
4. Neutron dose estimates and discussion

In order to consider the biological consequences of the neutrons, it is necessary to calculate the neutron dose based on the experimental fluence measurements and on the Monte Carlo simulations.

The neutron ambient dose equivalent, H*(10), has been calculated based on the obtained neutron fluence from measurements and Monte Carlo simulations and conversion factors from the ICRP 74 report (ICRP, 1996).

Furthermore, to be able to report the doses in clinical relevant units and to enable a better foundation for a comparison between treatment modalities we are required to consider how much radiation output is needed to deliver a prescribed dose to a target volume. For active proton and carbon ion therapy the radiation output may be quantified by the number of primary ions needed, while for photon therapy the concept of Monitor Units (MU) is used.

4.1 Normalization of neutron doses to treatment Gy

The amount of radiation output needed for delivering a prescribed dose was studied using the TRiP 98 (TReatment planning for Particles) code (Krämer and Scholz, 2000) for proton and carbon ion beams and the Varian Eclipse 11 treatment planning systems for photons.

The normalization study for particle therapy is based on 40 different treatment plans for carbon ions and protons with cubical target volumes ranging from 1 x 1 x 1 cm\(^3\) to 8 x 8 x 8 cm\(^3\) for 5 different phantom sizes. A similar study was performed for photons, but with a lower limitation in the volume size of 3 x 3 x 3 cm\(^3\). From the treatment planning systems the number of MUs, protons or carbon ions needed to deliver 1 Gy to the target volume for each treatment plan was obtained.
Results from the normalization calculations with the target volume centered at 8.6 cm depth are shown in Figure 4.1. We see that the number of primary ions per Gy increases steadily with the target volume size as expected. Only a small increase was observed when moving towards higher energies, i.e. targets at deeper positions. The number of MU remains more or less constant with increasing target size, but is expected to increase with increasing depths.

A reference volume of $3 \times 3 \times 3$ cm$^3$ at 8.6 cm depth in a water phantom was selected as an appropriate normalization setting for the experimental data in this work. The isocenter depth was 8.6 cm in the photon therapy experiments and this was also the Bragg peak depth for 107 MeV protons and 200 MeV/u carbon ions. In addition, small radiation fields were used in the experiments, and it is therefore natural to assume that the experimental results represent neutron doses associated to radiation treatment with small target volumes rather than neutron dose associated to radiation treatment with larger target volumes. More results from the normalization study are presented in Figure C-1 and Figure C-2 in the appendix.
Neutron dose estimates and discussion

The normalization study does not take into account the fact that IMRT and VMAT treatments usually require significantly more radiation output, i.e. Monitor Units (MU) than conventional treatment with open fields (Howell et al., 2005, Foroudi et al., 2012). The number of MUs needed per treatment Gray using IMRT or VMAT may be considerably higher than what was calculated in this normalization study. Furthermore, the RBE values for protons and carbon ions are not included. Morita et al. found RBE values between 1.5 and 3.5 in the SOBP from carbon ions (Morita et al., 1996). Therefore, the number of carbon ions needed to deliver one Gray equivalent (GyE) in the target volume may be significantly lower than the number of ions needed per Gy. However, it is not feasible to calculate the RBE value of carbon ions for the general case because the RBE depends on many factors. A clinical RBE value of 1.1 typically is used in proton therapy (Paganetti et al., 2002). It would therefore be possible to include the RBE for protons, but for the sake of consistency only the absorbed dose (Gy) is considered in the further data analysis.

4.2 Neutron dose estimates

In order to estimate the neutron dose from the fluence measurements with the SRAM detector, fluence-to-dose conversion factors were calculated for each measurement position based on the neutron energy spectrum in the respective positions. The energy spectra were obtained through Monte Carlo simulations. Because the neutron detection threshold for the SRAM detector is on the order of 1 MeV, the effective conversion factors have been calculated considering only neutrons with energies above 1 MeV.

4.2.1 Proton beams

In Figure 4.2 the neutron dose measured with the SRAM detector is shown as a function of lateral distance from the beam axis. In the panel to the left the dose is reported per primary proton, while the right panel reports dose per Gy, normalized to a 3 x 3 x 3 cm$^3$ target volume. The measured dose per Gy range from 0.6 mSv/Gy at
4 Neutron dose estimates and discussion

5.2 cm distance, to 0.2 mSv/Gy at 13.7 cm. On average, the measurements yielded about 50% higher neutron doses compared to the Monte Carlo simulations.

![Figure 4.2: The figures show the measured neutron dose per primary proton (left) and per Gray (right) as a function of lateral distance from the beam axis. The simulations show both dose from all neutrons (blue curve) and dose from neutrons above 1 MeV (red curve).](image)

Figure 4.3 shows the neutron dose as a function of depth. Again we see that the measurements yield higher neutron doses than the Monte Carlo simulations. The neutron doses remain relatively stable with depth. An increase in dose is seen in the measurements for more shallow depths at 5.2 cm lateral distance. This trend is not seen in the Monte Carlo simulations, and this may indicate that charged particles have contributed more to signal in the detector in these positions.

Figure 4.4 shows a two dimensional plot of the neutron dose obtained from Monte Carlo simulations of the proton beam experiment at TSL. We see that the maximum dose of about 10 mSv/Gy is found at the primary beam axis at depths before the Bragg peak, and the dose is reduced relatively fast both distal to the Bragg peak, and in the lateral direction.

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Neutron dose estimates and discussion

Figure 4.3: Neutron dose equivalent per primary proton (left) and per Gray (right) as a function of depth measured with the SRAM detector at three different lateral distances. The dotted and dashed lines represent FLUKA Monte Carlo simulations of the neutron dose for all neutron energies and for neutrons above 1 MeV, respectively.

Figure 4.4 – Monte Carlo simulations: Neutron dose equivalent from 178 MeV protons. The primary beam enters from the left side in the figure. The dose is reported in mSv per Gray normalized to a 3 x 3 x 3 cm³ target volume requiring 5.65 x 10⁹ protons per Gray.

The number of in-phantom neutron dose measurements from proton pencil beams are limited, but Schneider et al. performed measurements with CR-39 etch track detectors at the Bragg peak depth of a 177 MeV proton beam (Schneider et al., 2002). Neutron doses of 0.14 and 0.02 pSv per proton were found for lateral distances of 6.6 and 11.6
Neutron dose estimates and discussion

In Table 4.1 the lateral dose results from this work are summarized and compared to measurements from the study by Schneider et al. We see that there is relatively good agreement between the two experiments.

*Table 4.1: Summary of the measured neutron doses as a function of lateral distance to the beam axis. The findings are compared to neutron dose with measurements with CR-39 detectors (Schneider et al., 2002).*

<table>
<thead>
<tr>
<th>Lateral distance from beam axis [cm]</th>
<th>This work $H^*_10$ [pSv ion$^{-1}$]</th>
<th>(Schneider et al., 2002) $H^*_10$ [pSv ion$^{-1}$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.2</td>
<td>0.10±0.03</td>
<td>-</td>
</tr>
<tr>
<td>6.6</td>
<td>-</td>
<td>0.14</td>
</tr>
<tr>
<td>7.3</td>
<td>0.05±0.01</td>
<td>-</td>
</tr>
<tr>
<td>9.5</td>
<td>0.04±0.01</td>
<td>-</td>
</tr>
<tr>
<td>11.4 (11.6 in Schneider et al.)</td>
<td>0.02±0.01</td>
<td>0.02</td>
</tr>
<tr>
<td>13.7</td>
<td>0.02±0.01</td>
<td>-</td>
</tr>
</tbody>
</table>

The experiments with the SRAM detector in the proton beam indicate that the detector may be used for the purpose of neutron dose assessment in proton therapy. Further, the measurements and Monte Carlo simulations indicate that the neutron dose from 178 MeV protons is approximately 2 mSv/Gy at the Bragg peak and is reduced to approximately 0.1 mSv/Gy at 13.7 cm lateral distance from the beam axis.

4.2.2 Carbon ion beams

Due to the large uncertainties in the neutron fluence measurements in the carbon ion experiment with the water phantom, the estimates of neutron dose in carbon ion therapy are based on FLUKA Monte Carlo simulations, except for the measurements with the anthropomorphic phantom.

From the measured neutron fluence found from the measurements with the anthropomorphic phantom the ambient dose equivalent, $H^*_10$, can be calculated. The dose is calculated for the location of the detector which is 12 cm behind the phantom at 0°. The fluence was found to be $(1.6±0.4) \times 10^6$ n cm$^{-2}$ Gy$^{-1}$ where 1 Gy was equivalent to $4.32 \times 10^8$ carbon ions. An effective fluence-to-dose conversion
4 Neutron dose estimates and discussion

A factor of 295 pSv cm² was applied, based on the neutron energy spectrum obtained with Monte Carlo simulations. This gives the dose:

\[ H^*(10) = 1.6 \times 10^6 \text{cm}^{-2} \text{Gy}^{-1} \times 295 \text{ pSv cm}^2, \quad (4.1) \]

which gives:

\[ H^*(10) = (0.5 \pm 0.1) \text{ mSv Gy}^{-1}, \quad (4.2) \]

or in terms of dose per primary ion

\[ H^*(10) = (1.1 \pm 0.2) \text{ pSv Pr}^{-1}. \quad (4.3) \]

Dedicated FLUKA Monte Carlo simulations based on the delivered treatment plan indicated neutron dose of 0.5 mSv Gy⁻¹ in the measurement position. This is in good agreement with the measurements.

In a study by Gunzert Marx et al. (Gunzert-Marx et al., 2008), the neutron fluence at 3 meter distance from a water target irradiated by 200 MeV/u carbon ions was measured with a BaF₂ telescope. The neutron ambient dose equivalent was found to be 10⁻² pSv per carbon ion at 0°. Similar values were also found in (Iwase et al., 2007) using the WENDI-II neutron detector. Gunzert-Marx et.al estimated the dose to be 6.4 pSv per carbon ion at the distal end of the water phantom (4 cm behind the target volume). FLUKA Monte Carlo simulations in this thesis indicate that the neutron dose 4 cm behind the target is a factor of four higher than in the SRAM measurement position. The estimated dose from the SRAM detector measurements at this position is then 4.4±0.8 pSv per carbon ion which, considering the differences in experimental set-up, is in good agreement with the estimate from the study by Gunzert-Marx et.al. The estimated dose per carbon ion and per Gy is summarized in Table 4.2.
4 Neutron dose estimates and discussion

Table 4.2: Comparison of the estimated neutron dose 4 cm behind the target volume with a dose estimate from (Gunzert-Marx et al., 2008). The large difference in the neutron dose per Gy reflects that more ions were required per Gy for the treatment plan considered in the study by Gunzert-Marx et al.

<table>
<thead>
<tr>
<th></th>
<th>$H^*(10)$ [pSv ion$^{-1}$]</th>
<th>$H^*(10)$ [mSv Gy$^{-1}$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>This work</td>
<td>4.4±0.8</td>
<td>2.0±0.4</td>
</tr>
<tr>
<td>(Gunzert-Marx et al., 2008)</td>
<td>6.4</td>
<td>16</td>
</tr>
</tbody>
</table>

Monte Carlo simulations with carbon ions were performed under the same conditions as the 178 MeV proton beam experiment at TSL. In Figure 4.5 and Figure 4.6 the neutron dose from the Monte Carlo simulations is plotted as a function of lateral distance and depth, respectively. The dose is reported both per primary carbon ion and per treatment Gy based on a 3 x 3 x 3 cm$^3$ target volume. We see that the dose is on the order of 10 mSv per Gy at the beam axis and is reduced to about 0.1 mSv per Gy at 13.7 cm distance. The results indicate that neutrons with energies above 1 MeV on average account for about 95% of the dose. The dose varies little with depth for the presented positions (the experimental measurement positions).

Figure 4.5 - Monte Carlo simulations: Simulations of the neutron ambient dose equivalent from 341 MeV/u carbon ions as a function of lateral distance from the beam axis.
4 Neutron dose estimates and discussion

Figure 4.6 - Monte Carlo simulations: Neutron dose from 341 MeV/u carbon ions as a function of depth. The results indicate that the neutron dose is dominated by neutrons of energies above 1 MeV and changes little with depth.

Figure 4.7 shows the simulated distribution of neutron dose inside and outside the water phantom. The maximum dose of about 20 mSv per Gy is reached along the beam axis proximal to the Bragg peak, and the dose is reduced relatively fast both distal to the Bragg peak, and in the lateral direction.

Figure 4.7 – Monte Carlo simulations: Neutron dose from 341 MeV/u carbon ions. The primary beam enters from the left side in the figure. The dose is reported in mSv per Gray normalized to a 3 x 3 x 3 cm³ target volume requiring 3.26 x 10⁸ carbon ions per Gray.
4 Neutron dose estimates and discussion

The experiments with the SRAM detector in carbon ion beams indicate that the detector may be used for the purpose of neutron dose assessment in carbon ion therapy. Further, the Monte Carlo simulations indicate that the neutron dose from 341 MeV/u carbon ions is on the order of 10 mSv/Gy at the Bragg peak and is reduced to approximately 0.1 mSv/Gy at 13.7 cm lateral distance from the beam axis.

4.2.3 Photon beams

In Figure 4.8 the calculated neutron doses from bubble detector and TLD measurements are shown. The fast neutron dose is calculated on the basis of the measurements with the BDS neutron spectrometer, while the thermal neutron dose is calculated from the TLD measurements. The results indicate that the thermal neutron
4 Neutron dose estimates and discussion

dose is about an order of magnitude less than the fast neutron dose. The total dose ranges from $(2.9 \pm 0.6) \times 10^{-2}$ mSv per MU at the surface of the water phantom to $(1.4 \pm 0.3) \times 10^{-3}$ mSv per MU at 10 cm depth in water. The neutron dose is almost unchanged for different lateral positions.

In Figure 4.9 the neutron doses per Gy from both measurements and simulations are plotted. We see that the neutron dose ranges from $(4.9 \pm 1.0)$ mSv/Gy at the surface of the water phantom to $(0.3 \pm 0.1)$ mSv/Gy at 10 cm water depth the measurements while the simulation indicate neutron doses ranging from 0.76 mSv/Gy at the surface of the water phantom to 0.07 mSv/Gy at 10 cm depth in water. The doses are normalized to a $3 \times 3 \times 3$ cm$^3$ target volume, requiring 166 Monitor Units per Gy.

Figure 4.9: Measurements (solid lines) and Monte Carlo simulations (dashed lines) of neutron dose as a function of depth. The applied photon field was $3 \times 3$ cm$^2$ and the energy mode was 15 MV. The dose estimates are based on a normalization of 166 MU/Gy.

Figure 4.10 shows Monte Carlo simulations of neutron dose from a $3 \times 3$ cm$^2$ 15 MV photon field. The highest neutron dose is observed inside the treatment field at shallow depths in water. In this area the neutron dose is about 1 mSv per Gy (166 MU). The Monte Carlo simulation shows similar distribution of neutron fluence as the measurements, but the absolute values are significantly lower in the simulations.
4 Neutron dose estimates and discussion

Figure 4.10 - Monte Carlo simulations: Neutron ambient dose equivalent per treatment Gy (166 MU) from the Varian linac in the 15 MV energy mode. The maximum neutron dose of about 1 mSv/Gy is found inside and close to the treatment field. The phantom walls are indicated with the black lines and the applied 3 x 3 cm$^2$ photon field enters the phantom from the left side.

In Table 4.3 the results from this work is compared to measurements and simulations in the literature. In general it can be seen that at the surface of the water phantom the neutron doses reported are in the range from 5 µSv/MU to 37 µSv/MU while at 10 cm water depth the doses are typically reduced by a factor of ten or more.

The results presented in this section indicate that the neutron dose is highest at the surface and is reduced with depth. The neutron dose shows little dependence on the lateral distance from the treatment field. Further, the results indicate that the neutron dose is dominated by fast neutrons and that the dose from thermal neutron reaches a maximum at a few centimeters depth.

The results in this work indicate that neutron doses from 15 MV photons is in the range from approximately 0.8 mSv/Gy to 5 mSv/Gy for shallow depths and are about one order of magnitude lower at 10 cm depth.
4 Neutron dose estimates and discussion

Table 4.3: Comparison of the neutron dose equivalent obtained in this work with the literature. In the study by Martinez-Ovalle et al. the MCNPX Monte Carlo code was applied. In (Kry et al., 2005) neutrons were measured with gold foils while the study by Kaderka was performed with BDS bubble detectors.

<table>
<thead>
<tr>
<th>Measurements at phantom surface:</th>
<th>Linac model</th>
<th>Field size [cm²]</th>
<th>Energy [MV]</th>
<th>Neutron dose [µSv/MU]</th>
</tr>
</thead>
<tbody>
<tr>
<td>This work (BDS+TLD)</td>
<td>Varian 23 iX</td>
<td>3 x 3</td>
<td>15</td>
<td>29±6</td>
</tr>
<tr>
<td>This work (FLUKA)</td>
<td>Varian 23 iX</td>
<td>3 x 3</td>
<td>15</td>
<td>4.6</td>
</tr>
<tr>
<td>(Martinez-Ovalle et al., 2011)</td>
<td>Varian 2100C</td>
<td>10 x 10</td>
<td>15</td>
<td>37.0±0.9</td>
</tr>
<tr>
<td>(Kaderka, 2011)</td>
<td>Elekta SL25</td>
<td>5 x 5</td>
<td>18</td>
<td>18.8</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Measurements at 10 cm depth in phantom:</th>
<th>Linac model</th>
<th>Field size [cm²]</th>
<th>Energy [MV]</th>
<th>Neutron dose [µSv/MU]</th>
</tr>
</thead>
<tbody>
<tr>
<td>This work (BDS+TLD)</td>
<td>Varian 23 iX</td>
<td>3 x 3</td>
<td>15</td>
<td>1.4±0.3</td>
</tr>
<tr>
<td>This work (FLUKA)</td>
<td>Varian 23 iX</td>
<td>3 x 3</td>
<td>15</td>
<td>0.5</td>
</tr>
<tr>
<td>(Martinez-Ovalle et al., 2011)</td>
<td>Varian 2100C</td>
<td>10 x 10</td>
<td>15</td>
<td>2.7±0.1</td>
</tr>
<tr>
<td>(Kry et al., 2005)</td>
<td>Varian 2100</td>
<td>IMRT</td>
<td>15</td>
<td>3.2</td>
</tr>
<tr>
<td>(Kaderka, 2011)</td>
<td>Elekta SL25</td>
<td>5 x 5</td>
<td>18</td>
<td>1.2</td>
</tr>
</tbody>
</table>

4.3 Comparison of neutron doses in photon, proton and carbon ion therapy

The results presented in this section are based on the dose estimated presented in this chapter, and the normalization using a target volume of size 3 x 3 x 3 cm³ located at 8.6 cm depth. This means that $5.65 \times 10^9$ protons, $3.26 \times 10^8$ carbon ions or 166 Monitor Units are needed to deliver one Gray. The 166 MUs reflect the radiation output needed in photon therapy and corresponds to $6.96 \times 10^{14}$ primary electrons in the Monte Carlo simulations. The difference in neutron distribution from a single pencil beam compared to that from a 3 x 3 cm² scanned field is not considered.

Measurements from 178 MeV protons are plotted with and without energy and normalization corrections. The corrections are applied in order to scale the measured
4 Neutron dose estimates and discussion

neutron dose at 178 MeV to the expected detector response from 107 MeV protons. The correction factors are based on the difference in neutron dose for the two energies found in Monte Carlo simulations.

Figure 4.11 - Measurements and Monte Carlo simulations: Comparison of the neutron dose equivalent from photon, proton and carbon ion therapy per Gy absorbed dose in a 3 x 3 x 3 cm$^3$ target volume. The dose is reported for 8.6 cm depth$^1$ and is given as a function lateral distance from the beam axis. The neutron dose from carbon ions is reported per absorbed dose. Taking into account the higher RBE of carbon ions in the Bragg peak, the neutron doses from carbon and proton may be similar.

Figure 4.11 shows a comparison of the neutron dose from the three treatment modalities as a function of lateral distance from the primary beam axis. The plot indicates that the neutron dose from carbon ions is higher than for protons. Comparing the Monte Carlo simulations for protons and carbon ions indicates a factor of four higher neutron dose at the beam axis and typically a factor of 1.7 higher neutron dose at 3 cm or larger lateral distance. However, with the inclusion of the

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$^1$ The data for 178 MeV protons are reported for the Bragg peak depth of this energy (21 cm). Further, the measured neutron dose from proton therapy at 178 MeV is normalized with the number of protons needed to deliver one Gray at 20 cm depth to the same 3 x 3 x 3 cm$^3$ target volume (6.52 x 10$^9$ protons).
4 Neutron dose estimates and discussion

RBE similar neutron doses may be expected from protons and carbon ions lateral to the primary beam. The SRAM detector measurements also indicate lower neutron doses from protons than the values obtained from Monte Carlo simulations of carbon ions. Both Monte Carlo simulations and measurements indicate that the neutron dose from photons remains almost constant with respect to the lateral distance from the beam axis. Overall, the results demonstrate that the neutron dose from protons and carbon ions may be higher than the neutron dose from photons close to the primary beam axis. Further the results indicate that at a lateral distance of approximately five to ten centimeters, the neutron dose from photons may surpass the neutron dose from protons and carbon ions and remain higher for larger distances from the beam axis.

In Figure 4.12 the neutron dose is shown as a function of depth in water at 5.5 cm lateral distance from the beam axis. The plot illustrates the strong dependence of depth observed in neutron dose estimates from photon therapy. Both the Monte Carlo

![Neutron depth dose](image)

**Figure 4.12 - Measurements and Monte Carlo simulations: Comparison of the neutron dose equivalent from photon, proton and carbon ion therapy per Gy absorbed dose in a 3 x 3 x 3 cm³ target volume. The dose is given as a function depth in water for a lateral distance of 5.5 cm from the beam axis.**
4 Neutron dose estimates and discussion

simulations and measurements indicate that the neutron dose from photons is higher than the neutron dose from protons and carbon ions at shallow depths. While the neutron dose from photons is reduced with depth, a build-up of neutron dose is observed in particle therapy. At a depth perhaps between five and ten centimeters the neutron dose from protons and carbon ions may surpass the neutron dose from photons.

The results presented in this section indicate that neutron doses from the considered scenarios may range from about 10 mSv per Gy and down to three orders of magnitude less neutron dose for most peripheral positions. Larger treatment volumes will result in higher neutron doses in particle therapy because the number of primary ions needed increases approximately linearly with the treatment volume. In photon therapy, the neutron dose is strongly dependent on the radiation output from the linac. If extensive beam collimation is performed, the number of monitor units, and thus the neutron dose, may be expected to increase with the treatment volume size. Thus, this effect must be quantified through considering specific treatment plans.
5. Conclusion and outlook

The objective of this work is to investigate the neutron doses from radiation therapy with photons, protons and carbon ions.

Neutron dose measurements were performed with the SRAM detector inside and outside a water phantom irradiated by a 178 MeV proton beam. The measurements showed reasonable agreement with the Monte Carlo simulations considering the uncertainties in the experiment. This indicates that the SRAM detector may be used for the purpose of neutron dose measurements in proton therapy. For in-phantom measurements the SRAM detector may be a good alternative to passive detectors due to its compact design. More information both on the energy dependence of the SRAM detector’s response to neutrons and its response to charged particles will increase the reliability of the measurements.

The measurements and Monte Carlo simulations indicate that the neutron dose from 178 MeV protons is approximately 2 mSv/Gy at the Bragg peak and that the neutron dose is reduced to approximately 0.1 mSv/Gy at 13.7 cm lateral distance from the beam axis. Further, the measurements suggest that the neutron dose changes little with depth in the range from 2 cm prior to the Bragg peak to 3 cm beyond the Bragg peak. The Monte Carlo simulations support these findings. In addition, the Monte Carlo simulations indicate that the neutron dose increases with depth in a build-up region of about five centimeters. Beyond this build-up region, only moderate changes in the neutron fluence are observed before approaching the Bragg peak depth where the neutron fluence starts to decrease with depth.

The experiments with the SRAM detector in carbon ion beams indicate that the detector may be used for the purpose of neutron dose measurements in carbon ion therapy. The Monte Carlo simulations suggest that the neutron dose from 341 MeV/u carbon ions are on the order of 10 mSv/Gy at the Bragg peak. The neutron dose is reduced to approximately 0.1 mSv/Gy at 13.7 cm lateral distance from the beam axis.
5 Conclusion and outlook

The Monte Carlo simulations indicate therefore that the neutron dose from carbon ions is higher close to the primary beam compared to the neutron doses originating from protons, while at greater lateral distances from the beam axis the neutron doses are comparable. Taking into account the high RBE of carbon ions relative to the protons’ RBE, the neutron dose from proton therapy and carbon ion therapy may be expected to be similar.

The neutron dose from 15 MV photons was investigated in the area around the treatment field from a Varian 23 iX linear accelerator. The measurements in this part of the work indicate that neutron doses outside the treatment field range from 5 mSv/Gy at the surface of a water phantom to 0.2 mSv/Gy at 10 cm depth in water. The measurements indicate that the dose from neutrons is less than 1 mSv per treatment Gray for positions deeper than 5 cm in the water phantom. The Monte Carlo simulations suggest neutron doses outside the treatment field in the range between 0.8 mSv/Gy at the surface of the water phantom to 0.1 mSv/Gy at 10 cm depth in water. Both measurements and Monte Carlo simulations show that the neutron dose in photon therapy is strongly dependent on the depth in the patient, and the neutron dose is only to a low degree dependent on the lateral distance from the treatment field. Further, both measurement and Monte Carlo simulations indicate that the neutron dose is dominated by fast neutrons and that the dose from thermal neutron reaches a maximum at a few centimeters depth.

Moreover, the Monte Carlo simulations suggest that neutrons in photon therapy primarily are produced in the treatment head components. The Monte Carlo simulations also indicate that a large fraction of the neutron dose originates from the components closest to the patient, although neutron production may be higher in other components such as the primary collimator.

A comparison of the neutron dose at 8.6 cm depth from the different treatment modalities indicate that close to the beam axis, the neutron dose from protons and ions may be higher than the neutron dose from photons. However, at a lateral distance
of approximately five to ten centimeters, the neutron dose from photons may surpass the neutron dose from protons and carbon ions and remain higher for greater distances. Further, both Monte Carlo simulations and measurements at 5.5 centimeters distance from the beam axis indicate that the neutron dose from photons is higher than the neutron dose from protons and carbon ions at shallow depths. At a depth between five and ten centimeters the neutron dose from protons and carbon ions may surpass the neutron dose from photons.

The assessment of neutron dose from radiation therapy is a topic which is still in need of further research effort. The potential for improvements on the detector side is clearly present, as well as the need for additional measurement data. The results from this work can be used as a platform from which to further investigate and quantify the neutron field around particle therapy beams and medical linear accelerators. A prolonged research effort is needed to identify the neutron doses to patients receiving radiation treatment, and as an extension of this, enable an inclusion of the neutron dose component and its biological effect in clinical treatment planning software tools.
5 Conclusion and outlook
References


BEDOJNI, R. 2006. *Neutron spectrometry and dosimetry for radiation protection around a high energy electron/positron collider* Ph.D., Universitat Autònoma de Barcelona.

BEIR 2006. *BEIR VII, Health risks from exposure to low levels of ionizing radiation*.


MORITA, K., H.TUJII & K.KAWACHI. 1996. Therapy at HIMAC.


References


References


Appendix A  Detector data and experimental results

In Table A-1 technical information on the SRAM chips used for the SRAM neutron detector is shown, and Table A-2 summarizes the assumed neutron cross section for the SRAM detector.

*Table A-1: Technical specifications of the SRAM chips used in the SRAM detector.*

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>150</td>
<td>55</td>
<td>16</td>
<td>2.43</td>
</tr>
</tbody>
</table>

*Table A-2: The energy dependence of the SRAM detector. The results are based on radiation tests at PTB Braunschweig, the Oslo Cyclotron Laboratory, The Svedberg Laboratory (Uppsala), The Institute for Energy Technology (Kjeller) and CERF (the Cern-EU high-energy reference field).*

<table>
<thead>
<tr>
<th>Energy[MeV]</th>
<th>Absolute SEU cross section [SEU cm$^2$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thermal</td>
<td>$(8 \pm 2 ) \times 10^{-8}$</td>
</tr>
<tr>
<td>5.8</td>
<td>$(6.2 \pm 0.9) \times 10^{-7}$</td>
</tr>
<tr>
<td>8.5</td>
<td>$(1.3 \pm 0.1) \times 10^{-6}$</td>
</tr>
<tr>
<td>14.8</td>
<td>$(4.0 \pm 0.4) \times 10^{-6}$</td>
</tr>
<tr>
<td>26</td>
<td>$(4.5 \pm 0.5) \times 10^{-6}$</td>
</tr>
<tr>
<td>20-200</td>
<td>$(9.3 \pm 1.9) \times 10^{-6}$</td>
</tr>
<tr>
<td>175</td>
<td>$(9.9 \pm 1.1) \times 10^{-6}$</td>
</tr>
</tbody>
</table>

Table A-3 shows the obtained values from neutron cross section measurements with the BDT detectors in thermal, 5.8 MeV, 8.5 MeV, and 14.8 MeV neutron beams.

1 The interval 20-200 MeV refers to the mixed radiation field at CERF and calculations where the contribution from lower energy particles has been subtracted.

2 The SEU cross section for thermal neutrons is defined only as an upper limit of the cross section, indicating that the actual cross section may be lower.
Appendix A Detector data and experimental results

Table A-3: Summary of the cross section measurements for the BDT bubble detector.

<table>
<thead>
<tr>
<th>Neutron energy [MeV]:</th>
<th>Thermal</th>
<th>5.8</th>
<th>8.5</th>
<th>14.8</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cross section</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>[10^{-5} b/(n.cm^{-2})]</td>
<td>3.9±1.0</td>
<td>9.2±2.8</td>
<td>7.8±2.4</td>
<td>7.7±2.9</td>
</tr>
<tr>
<td>Dose response</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>[b/µSv]</td>
<td>3.7±0.9</td>
<td>0.23±0.07</td>
<td>0.19±0.06</td>
<td>0.14±0.05</td>
</tr>
</tbody>
</table>

BDS 2500 and BDS 10 000 bubble detectors were irradiated by a 174.5 MeV neutron beam at TSL. The purpose of this experiment was to investigate the bubble detectors cross section (sensitivity) to neutrons in this energy range. To ensure the correct temperature for the detectors during measurements these were insulated in a custom Styrofoam holder. Although the Styrofoam may to some degree moderate the neutron beam the effect is assumed to be small for this set-up. The results from the irradiation are summarized in Table A-4 and show values similar to the cross sections at lower energies (see Figure 2.4). The results indicate that bubble detectors have a potential for measurements at neutron energies above the initial measurement range intended by the vendor (10 keV – 20 MeV).

Table A-4: The table shows the results of the bubble detector irradiations in the 174.5 MeV neutron beam. 6 detectors of each type were irradiated.

<table>
<thead>
<tr>
<th>Detector</th>
<th>Cross section [bubbles/(n cm^{-2})]</th>
</tr>
</thead>
<tbody>
<tr>
<td>BDS 2500</td>
<td>(1.2 ± 0.1) x 10^{-4}</td>
</tr>
<tr>
<td>BDS 10 000</td>
<td>(3.3 ± 0.4) x 10^{-5}</td>
</tr>
</tbody>
</table>

In Figure A-1 neutron energy spectra obtained with the BDS detectors are shown. Only small changes in the spectra were observed for different lateral distances from the photon field edge at the same depth in water. As the figure shows, the neutron fluence is strongly dependent on the depth in water.
Appendix A Detector data and experimental results

Figure A-1: Neutron energy spectra obtained with the BDS detectors and the BUNTO unfolding code. The results show that the spectra are almost unchanged in shape and amplitude for measurement positions at the same depth with different lateral distances from the field edge. The error for each bin is assumed to be 20%.
Appendix A Detector data and experimental results
Appendix B  Monte Carlo simulations

Figure B-1 shows the neutron production rate, the neutron dose contribution inside the treatment field, and the neutron dose contribution outside the treatment field for the major components of the linear accelerator studied in Monte Carlo simulations in this work.

Figure B-1: The plots show a comparison of each region’s neutron production and the region’s contribution to neutron dose in a water phantom. While neutron production is largest in the primary collimator, neutrons from other components such as the jaws and MLCs may contribute more to the neutron dose in the patient.
Table B-1 shows the obtained neutron production for the major components of the Varian linear accelerator during Monte Carlo simulations with four different collimator settings.

*Table B-1: An overview of the neutron production region by region for different collimator settings. The full collimator settings are described in Table 2.1.*

<table>
<thead>
<tr>
<th>Collimator setting:</th>
<th>1 (3x3)</th>
<th>2 (10x10)</th>
<th>3 (MLCs 10 cm)</th>
<th>4 (MLCs closed)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Primary collimator</td>
<td>47.5%</td>
<td>49.2%</td>
<td>50.5%</td>
<td>46.9%</td>
</tr>
<tr>
<td>Target (tungsten)</td>
<td>4.0%</td>
<td>4.1%</td>
<td>4.2%</td>
<td>3.9%</td>
</tr>
<tr>
<td>Target (copper)</td>
<td>6.4%</td>
<td>6.6%</td>
<td>6.8%</td>
<td>6.3%</td>
</tr>
<tr>
<td>Flattening filter</td>
<td>1.8%</td>
<td>1.9%</td>
<td>1.8%</td>
<td>1.7%</td>
</tr>
<tr>
<td>Y-Jaws</td>
<td>31.6%</td>
<td>25.1%</td>
<td>16.5%</td>
<td>15.3%</td>
</tr>
<tr>
<td>X-Jaws</td>
<td>8.4%</td>
<td>12.5%</td>
<td>12.2%</td>
<td>11.4%</td>
</tr>
<tr>
<td>MLCs</td>
<td>0.03%</td>
<td>0.02%</td>
<td>7.2%</td>
<td>14.2%</td>
</tr>
<tr>
<td>Remaining components</td>
<td>0.3%</td>
<td>0.6%</td>
<td>0.8%</td>
<td>0.2%</td>
</tr>
</tbody>
</table>
Appendix C  Dose estimates and Normalization study results

Figure C-1 shows the number of primary protons and carbon ions needed to deliver 1 Gy to a 3 x 3 x 3 cm³ target volume as a function of the target volumes position in depth in a water phantom. The results were obtained with the TRiP 98 (TReatment planning for Particles) code.

![Graph showing calculated radiation output needed per Gy](image)

*Figure C-1: The number of primary protons and carbon ions needed to deliver 1 Gy to a 3 x 3 x 3 cm³ target volume. In the blue curve the number of protons is divided by 20 in order to better illustrate the difference between the number of protons and the number of carbon ions needed. The results were obtained with the TRiP 98 (TReatment planning for Particles) code (Krämer and Scholz, 2000).*

Figure C-2 shows the number of primary protons and carbon ions needed to deliver 1 Gy to a target volume in a water phantom as a function of the target volume size. The results were obtained with the TRiP 98 (TReatment planning for Particles) code.
Figure C-2: The number of primary protons and carbon ions needed to deliver 1 Gy as a function of target volume size. The target volumes were all defined in the center of a 20 x 20 x 20 cm³ water phantom. In the blue curve the number of protons is divided by 20 in order to better illustrate the difference between the number of protons and the number of carbon ions needed. The results were obtained with the TRiP 98 (TReatment planning for Particles) code (Krämer and Scholz, 2000).