

**NPL REPORT
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LIGHT-ION BEAM DOSIMETRY

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Thomas, David Shipley,
Andrzej Kacperek**

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LIGHT-ION BEAM DOSIMETRY

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ABSTRACT

In this report an overview is given of the scientific work done in the Strategic Research project “Improved light ion beam dosimetry” and work under the 2001-2004 National Measurement System, Ionising Radiation Metrology Programme of the UK Department of Trade and Industry. It is amended with work performed under the 2004-2007 National Measurement System, Ionising Radiation Metrology Programme of the UK Department of Trade and Industry to finalize work initiated in the earlier programmes. This work also substantially benefited from research projects from Clatterbridge Centre of Oncology and from the Natural Sciences and Engineering Research Council of Canada (NSERC) and collaboration with Ghent University Belgium.

The introduction is a very brief initiation in the field of proton beam radiotherapy and the second chapter provides an extensive overview of the literature. The third chapter presents and discusses the experimental work performed in the scope of this project and the fourth chapter discusses the simulations that were done in support. The conclusions section summarises the key achievements and the opportunities this work created for NPL.

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1. INTRODUCTION

Radiotherapy with proton beams is a superior cancer treatment modality over x-ray and electron beam treatment. Radiotherapy with proton beams is half a century old but was much slower to become established than x-ray and electron beam treatment, mainly due to the higher cost of the facility involved. Nowadays, a full commercial proton therapy facility with three patient gantries and two ocular beam lines costs around £40M (Goitein and Jermann 2003). In such a facility, the cost per treatment fraction including operational and capital cost is a factor of three higher than for conformal x-ray therapy and according to the most optimistic estimates this ratio can be potentially reduced to a factor of two (Goitein and Jermann 2003). The superiority of protons lies in the fact that the radiation dose can be confined largely to the tumour with much lower doses to the surrounding healthy tissue than can be achieved with x-rays. This allows for dose escalation in the tumour itself and therefore better control of the treatment outcome.

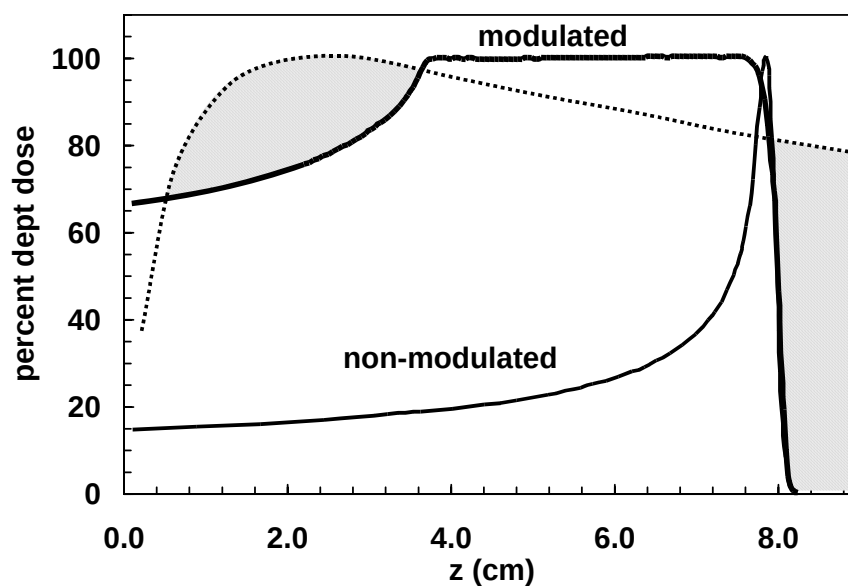


Figure 1. Percent depth dose profiles for 100 MeV protons in water (solid lines). For comparison, the dashed line represents the depth dose distribution for a 10 MV x-ray beam and the shaded areas indicate the regions where a modulated proton beam offers a physical advantage in aiming to spare healthy tissue.

The reason for the superior potential of a proton beam lies in its characteristic depth dose distribution, exhibiting a typical Bragg peak (BP) near the maximal penetration depth, as shown in figure 1, and steep penumbrae. Light-ion beams have the additional advantage of having a higher relative biological effectiveness (RBE) compared to x-rays and electrons. Energy or range modulation of the beam allows the generation of a flat dose profile over a certain interval of depth, the so-called spread-out Bragg peak (SOBP). Spot scanning offers a different approach for conformal irradiation of the tumour volume. One has to distinguish between two energy ranges that are used for proton therapy. Proton beams with energies lower than 70 MeV penetrate just a few centimeters in tissue and are mainly used for the treatment of ocular melanoma of the retina and the iris. For retina melanoma, this treatment is mainly considered when the tumour is close to the optical nerve or the macula and result in very high cure rates (above 99% with retention of the eye) (Egger et al. 2003). High-

energy protons (above 150 MeV) can be used for any type of tumour and will be advisable whenever there are critical tissue structures or organs close to the tumour. This form of treatment is only delivered at a serious scale since one decade and although the first clinical trials emerging from these applications show an improvement with respect to morbidity compared to x-ray therapy (Slater et al. 2004), large scale clinical trials are not as yet available.

Despite its slow introduction in radiotherapy, the number of treatments with proton beams, and even heavier ions, is now increasing rapidly. About 40,000 patients have been treated with light ions so far. Many oncology departments worldwide are now in the process of getting a proton beam and there are now six commercial companies involved in the production of proton therapy dedicated facilities. At present, there are five operational proton therapy facilities in the US and another two under construction, Japan has six proton facilities and another one under construction and in Europe there are at present nine proton facilities, among which one in the UK, and another four under construction. Remarkable is also that China and Korea have recently started investing in several facilities. Light ion beams (mainly ^{12}C ions) are used for their advanced biological effect compared to protons and at present two facilities in Japan and one in Germany are able to treat patients. There is a tendency, however, to integrate the possibility of treating with protons and light ions in one facility using the same cyclotron. Figure 2 illustrates the evolution of the number of light-ion therapy facilities worldwide. The black bars represent the operational facilities; the grey bars those that will start treatments in the next 8 years. For more information on the development of proton and ion therapy centres worldwide, (see Sisterton 2005).

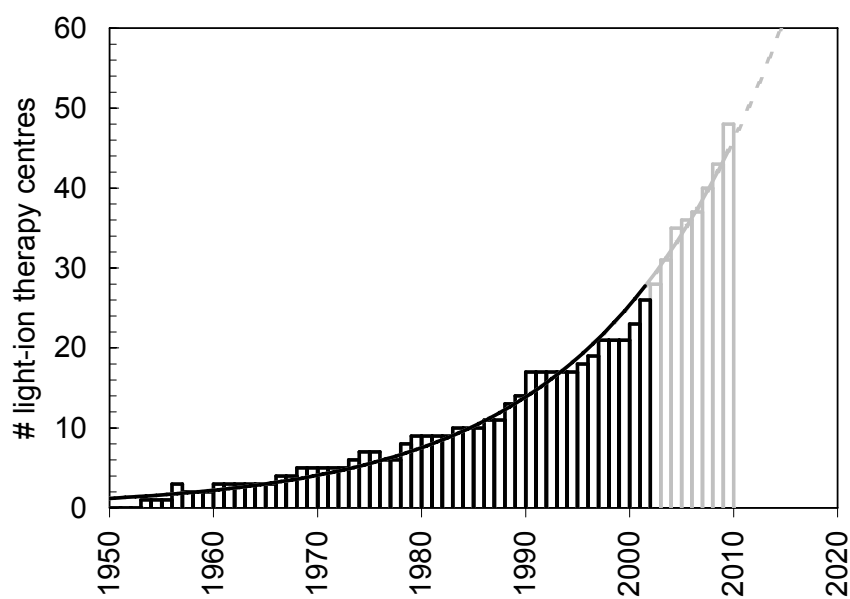


Figure 2. Number of light-ion therapy centres in operation (black), and planned in the next 8 years (grey)

In the UK, treatment of ocular tumours has been carried out successfully since 1989 with a 62 MeV proton beam at Clatterbridge. Several proposals have been made recently for a high-energy facility in the UK but without success so far. The proposed CASIM (Centre for Accelerator Science, Imaging and Medicine) facility at Daresbury, intended to offer 230 MeV proton beams and also heavier ions, failed to get governmental approval. Proposals to

increase the energy of the Clatterbridge beam line using a linear accelerator booster and to build a dedicated proton/ion therapy facility in the London area (Jones 2005) are pending.

Dosimetry in proton beams is not as well established as in x-ray and electron beams. Modern dosimetry recommendations quote calorimetry as the preferred primary dose standard for proton beams, but no such standard exists and only a few isolated calorimetry experiments in proton beams have been performed. Good progress has been made in the improvement of ionisation chamber dosimetry, but inconsistencies between dosimetry by different centres of 3% are still not uncommon. Near the end of the proton range (in the so called Bragg peak), spectral characteristics of protons are quite different compared to the entrance region. It is well known that the Relative Biological Effectiveness (the ability to kill cells) in the Bragg peak, and particularly in the distal side of it, is enhanced. Usually, the Bragg peak is spread out to cover the tumour in modulated beams. For ocular tumours, which can be very small, the Bragg peak has to be positioned precisely and the shape of the Bragg peak has to be known exactly. The ideal dosimeter for Bragg peak measurements has not yet been established. The dosimetric and spectral influence of beam shaping devices such as wedges, collimators and modulator wheels is not known precisely.

Monte Carlo simulation of photon and electron transport in radiotherapy physics has become a well-established research technique. For light-ion beam treatment, transport codes are available but have not been used extensively in the field of radiotherapy. Especially new codes such as GEANT, MCNPX and PTRAN have not been used much in this respect.

The SR project 'Improved light ion dosimetry' was in the first place intended to initiate and enhance NPL's skills in the field of proton dosimetry and to explore potential routes for improvement of the dosimetry chain.

The objectives of the project were:

- A literature survey on dosimetric methods and protocols in proton beams, more specifically on the use of ion chambers, diodes, alanine, alanine film and possibly radiographic film and gel dosimetry, and to apply and test these methods in the 60 MeV beam in Clatterbridge.
- The measurement of depth dose distributions with attention focussed on accurate measurements of the Bragg peak.
- The measurement of lateral dose profiles.
- Investigate the influence of beam shaping devices
- The measurement and calculation of detector perturbation.
- Support the measurements with Monte Carlo simulations using MCNPX, GEANT4 and PTRAN by calculation of three dimensional dose distributions (comparing the results obtained using the three codes), calculation of particle spectra, calculation of stopping power ratios and calculation of dose perturbation factors for detectors.

2. LITERATURE STUDY

An extensive reference list is established and maintained by the author. A copy can be obtained on request. In this chapter a literature overview on clinical light-ion beam dosimetry is given, which is intended as an introduction to those who have a general background in dosimetry but are newly working in the field of proton dosimetry. It is an up to date review of all the work published in the literature apart from the work done in the scope of NPL's projects over the last two years, which will be described in more detail in the succeeding chapters.

The literature review can be divided in three main parts: protocols and data, reference dosimetry and relative dosimetry. The first and second parts are partly a summary of a presentation given at the Montreal workshop "Recent Developments in Accurate Radiation Dosimetry" (Palmans and Vynckier 2002) and partly a more thorough discussion. The third part discusses various techniques used for relative dosimetry. Some of them, such as alanine, are investigated as secondary absolute dosimeters as well.

Previous reviews on the techniques used to deliver radiotherapy with proton beams, specific practical problems encountered in establishing a treatment and biological aspects are published by Chu et al. (1993), Bonnet (1993), Miller (1995) and Raju (1995). Chu et al. (1993) also give a thorough review of dosimetry techniques used. Bonnet (1993) and Miller (1995) devote some brief section on dosimetry. Further discussions of dosimetry methods have been reviewed to some extent in AAPM report 16 (Lyman et al. 1986), the ECHED Code of Practice (Vynckier et al. 1991, 1994) and ICRU report 59 (1998). Reference dosimetry has also been reviewed by Laitano (1998), Jones (2001), Palmans and Vynckier (2002) and Vynckier (2004).

2.1. History of light-ion beam dosimetry

Prior to 1979, there were only a few scientific publications dealing with absolute dosimetry for proton beams for clinical or pre-clinical use (for example by Boles et al. 1969, Momeni et al. 1973 and McDonald et al. 1977) and no systematic efforts in uniformising reference dosimetry for these types of beams have been reported. In general Faraday cups, calorimeters, chemical dosimeters or ionisation chambers were used for absolute dosimetry. For relative dosimetry various detectors have been used: thermoluminescent detectors, scintillators, radiographic and radiochromic film, semiconductor devices, chemical dosimeters.

Verhey et al. (1979) were the first to make a systematic comparison of different absolute dosimetry methods: Faraday cup dosimetry, calorimetry and ionisation chamber dosimetry. Since then, a number of protocols for reference dosimetry in clinical proton beams have been developed: AAPM Report-16 from TG-20 (Lyman et al. 1986), the ECHED protocol (Vynckier et al. 1991, 1994) ICRU report 59 (ICRU 1998) and IAEA TRS-398 (Andreo et al. 2000a). They all recommend one or more of the three methods studied by Verhey et al. (1979). Parallel with the developments of these protocols, new comprehensive sets of stopping power data (Janni 1982, ICRU report 49 1993, Medin and Andreo 1997) and nuclear interaction cross-section data (Janni 1982, ICRU report 63, 2000) have been published.

For radiotherapeutical heavy ions, the amount of literature is very limited and only recently a number of systematic dosimetry studies have been performed (Fukumura et al. 1998,

Hartmann et al. 1999, Jäkel et al. 2004, Kanai et al. 2004). IAEA TRS-398 contains the first international recommendation for the dosimetry of heavy ions.

2.2. Overview of dosimetry recommendations and Codes of Practice using ionisation chambers

2.2.1. Codes of Practice

The paper of Verhey et al. (1979) could be regarded as the first comprehensive recommendation and laid the basis for Report 16 of AAPM Task Group 20 (Lyman et al. 1986), the first Code of Practice for absolute dosimetry in clinical proton beams. In this report, the quantity of interest is defined to be dose to tissue. It recommends A-150 calorimetry as the preferred dosimetry method, since A-150 is a tissue equivalent (TE) plastic. The second preferred method is Faraday cup dosimetry. Both methods will be described and discussed in a later section. In practice, an A-150 walled ionisation chamber should be calibrated against one of the previous instruments and used for routine dosimetry. For a modulated beam, the reference point for the measurement of absorbed dose is the centre of the SOBP, in which case the dimensions of the ionisation chamber should be small compared to the extension along the beam axis of the SOBP. For a non-modulated beam, absorbed dose should be measured in the entrance plateau region, provided the energy of the beam is high enough. If no calorimeter or Faraday cup is available, an A-150 walled ionisation chamber calibrated in ^{60}Co could be used. The formalism for calculating the dose from the measured ionisation in the chamber and the calibration factors is then very similar to the formalism used for high-energy photon and electron dosimetry, with the exception that the average energy required to produce an ion pair in dry air for protons, $(w_{\text{air}})_p$, and for the electron spectrum in a ^{60}Co beam, $(W_{\text{air}})_{^{60}\text{Co}}$, need to occur explicitly in the equations, since they are not equal. AAPM report 16 gives only generic data for chambers with a TE wall (A-150) without taking into account any proton energy dependence (except for a distinction made between the SOBP region and the plateau region) of the required quantities. It also assumes that the ion chambers' build-up cap during calibration is made of A-150 as well. The standard uncertainty of dosimetry is estimated to be 5%, dominated mainly by the uncertainty on $(w_{\text{air}})_p$.

Five years later, a work group of the European Heavy Charged Particle Dosimetry Group (ECHED) published its own Code of Practice (Vynckier et al. 1991) which was mainly based on the same recommendations. The major improvements were that it lists data for a wider range of ionisation chambers and that the energy dependence of the quantities related with the proton beam is taken into account by introducing the effective proton energy E_{eff} as a beam quality specifier. Another difference is that water was proposed as the recommended phantom material for the absolute dose measurement, which was anticipating to a consensus that absorbed dose should be specified in water and not in tissue. Finally, an important difference is the use of another value of the average energy required for a proton to produce an ion pair in air, since the value in AAPM report 16 was based on a misquotation. The new value was also supported by more recent experimental work. A detailed discussion of the standard uncertainty was given, which was estimated to be 4%. Three years later a supplement to the ECHED protocol was published (Vynckier et al. 1994) mainly intended to implement the more recent stopping powers of ICRU report 49 (1993). Additionally, in accordance with recommendations for other beam types, the reference medium to which absorbed dose should be specified was changed into water. The possibility to use other ionisation chamber types not listed in the protocol is allowed and the required data should then be taken from given international references.

More recently another air kerma based formalism was published in ICRU report 59 (1999). However, Medin et al. (2000) pointed out that it contains some major inaccuracies, mainly regarding the data for ionisation chambers with a mixed chamber/build-up cap composition. Moreover, the treatment of the humidity correction was proven to be erroneous and the value for $(w_{\text{air}})_p$ that was proposed was not obtained by an objective method.

During the nineties, a number of papers were published (Vatnitsky et al. 1993, Medin et al. 1995, Vatnitsky et al. 1996a) which suggested to adopt a similar procedures based on absorbed dose to water as were then recently proposed for high-energy photon beams (Andreo 1992, Rogers 1992). These led to absorbed dose based formalisms in ICRU report 59 (1999) and IAEA TRS-398 (Andreo et al. 2000a). Again, Medin et al. (2000) showed that a major inaccuracy was introduced in ICRU report 59 by ignoring perturbation corrections for ionisation chambers in the ^{60}Co calibration beam and the absorbed dose based Code of Practice suffers in the same way from the choice of the $(w_{\text{air}})_p$ -value (which will be discussed below). The standard uncertainty is estimated between 2.5% and 3%.

IAEA TRS-398 finally presents a CoP for a wide range of ionisation chambers. A difference compared with the previous reports is that it explicitly quotes the calibration of an ionisation chamber in a proton beam as the preferred method. The data given are limited to a calibration in ^{60}Co as reference beam, since no primary standards for proton beams are available. Compared to previous CoP's, there are some substantial changes: (i) it uses the residual range R_{res} as beam quality specifier, (ii) Spencer-Attix water-to-air stopping power ratios taken from Medin and Andreo (1997) are used in stead of Bragg-Gray stopping power ratios, (iii) a rigorous numerical determination of the $(w_{\text{air}})_p$ -value was performed and (iv) data for plane-parallel chambers are given. A detailed uncertainty budget is given and the overall uncertainty of absorbed dose-to-water is estimated to be 2.0 % when using a cylindrical ionisation chamber and 2.3 % when using a plane-parallel ionisation chamber. Another interesting point is that a perturbation correction factor for the proton beam is introduced, although it is assumed to be unity for all ionisation chambers and all beam qualities that are listed in the protocol.

2.2.2. Data required for ionisation chamber dosimetry in clinical proton beams

2.2.2.1. Stopping powers

Until 1993, the most comprehensive set of stopping power data were those tabulated by Janni (1982). These were used in AAPM report 16 and by Vynckier et al. (1991). In addition, data calculated by Bichsel and Porter (1982) for the A-150 to tissue equivalent (TE) gas stopping power ratios was used in AAPM report 16. All later dosimetry recommendations use the more recent stopping power data of ICRU report 49 (1993). These offer the advantage of being more consistent with the ICRU report 37 (1984) electron stopping power data that are involved in ^{60}Co calibrations since they use the same values for the mean excitation energies. Medin and Andreo (1992) showed that discrepancies in the stopping powers of Janni and ICRU report 49 could be as large as 4% for the low-Z constituents involved in the ionisation chamber dosimetry protocols, mainly due to the different mean excitation energies used. A discussion between Ziegler (1999) and Seltzer et al. (2001) comment on the rationale of selecting certain data sets in ICRU report 49 (1993).

Medin and Andreo (1997) performed Monte Carlo simulations to calculate water-to-air Spencer-Attix stopping power ratios taking into account secondary protons and electrons. They showed that the influence of secondary protons on the water-to-air mass stopping power ratios is smaller than 0.1% for the energy range of clinical proton beams. The

inclusion of secondary electrons in the Spencer-Attix stopping power ratios, on the other hand, induces differences of up to 0.6% compared with the Bragg-Gray stopping power ratios. In IAEA TRS-398 these Spencer-Attix water-to-air stopping power ratios are implemented. The results of Medin and Andreo (1997) were confirmed by independent Monte Carlo simulations performed by Laitano and Rosetti (2000) using the FLUKA Monte Carlo code.

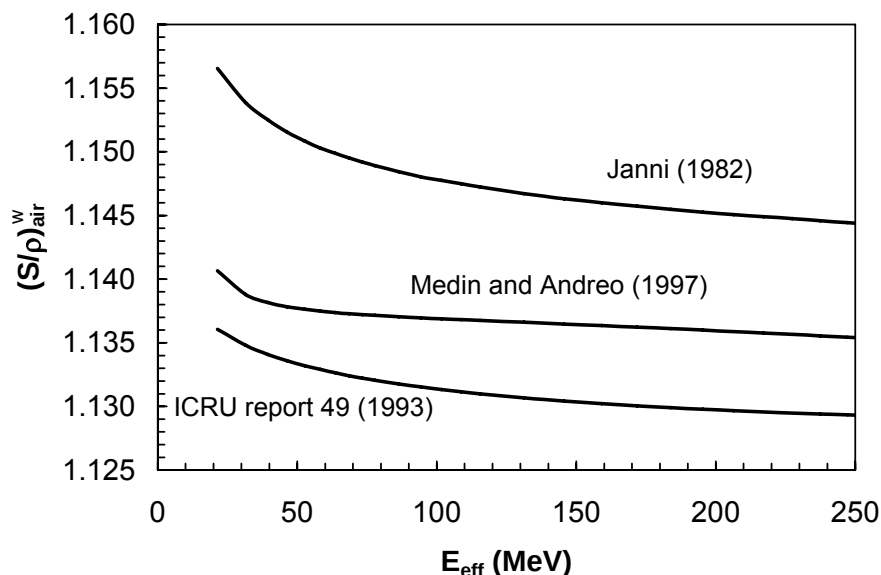


Figure 3. Water to air mass stopping power ratios for protons as a function of effective energy according to different sources. The values of Medin and Andreo (1997) are restricted stopping power ratios calculated by Monte Carlo simulations.

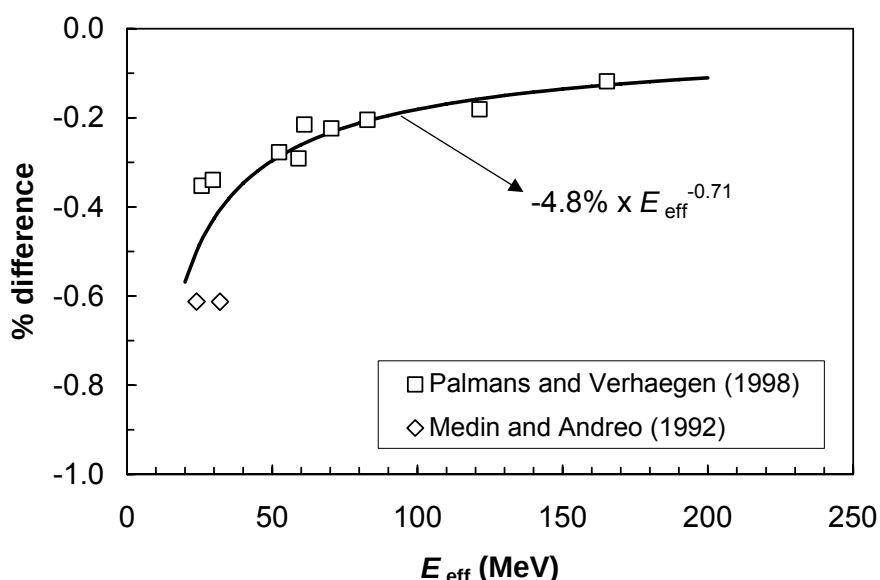


Figure 4. Percent difference of the water to air stopping power ratios calculated from the effective energy and calculated by integration over the proton spectrum.

Figure 3 compares the water-to-air stopping power ratios of Janni (1982), ICRU report 49 (1993) and Medin and Andreo (1997) as a function of energy for mono-energetic proton beams.

The proton stopping power ratios are in general selected as a function of a beam quality specifier, which can be the residual range R_{res} , defined as the distance of the point of measurement to the depth distal to the Bragg peak where the dose is reduced to 10% of the maximum dose, or the effective energy E_{eff} . Both are linked by the range-energy relationship for the stopping power data used and thus assume mono-energetic protons with the same range as the real beam. Seen the slow variation of the stopping power ratios with energy this is a good approximation for non-modulated proton beams, but in modulated beams the energy spectrum is generally wide and in principle the stopping powers should be integrated over the spectrum. Monte Carlo calculations were performed to evaluate the error induced due to this simplifying assumption for modulated beams by Medin and Andreo (1992) using LAHET and by Palmans and Verhaegen (1998) using PTRAN. The results of both investigations are summarized in figure 4 and show that the errors are limited to about 0.5% for effective energies down to 20 MeV. This lower energy corresponds with a residual range of 4mm.

2.2.2.2. Average energy required to produce an ion pair in dry air

It is very remarkable, that every of the four international reports (AAMP report 16, ECHED code of practice, ICRU report 59 and IAEA TRS-398) uses a different value of the average energy required to produce an ion pair in dry air. Since this value directly affect the outcome of a dose measurement with an ionisation chamber, the reason for these four different values deserves a discussion. Various notations have been used in different CoP's and some use the integral value $(W_{\text{air}})_p$, whereas others used the differential value $(w_{\text{air}})_p$, which could be justified given the small variation of this quantity above 20 MeV (Grosswendt and Baek, 1998). For the remainder of this section we will only use the notation $(w_{\text{air}})_p$ for all CoP's.

Table 1 gives an overview of experimental values available and contains all the information that led to the different values, except for the one used in AAPM report 16. The latter recommended a value of 34.3 eV referring to Verhey et al. (1979) which was, however, based on a misquotation of the value given by Myers (1968) as pointed out by Hayakawa and Schechtman (1988). In a reply, Verhey and Lyman (1992) defended the value of AAPM report 16 by noting that the value of Larson (1958) was determined for an energy of only 1.83 MeV and arguing that $(w_{\text{air}})_p$ should decrease considerably when increasing the energy from 2 MeV to clinical beam energies. They used the semi-empirical model of Dennis (1972, 1973), which was based on the assumption that for high proton energies, $(w_{\text{air}})_p$ should evolve asymptotically to the value for electrons. This model predicts that when increasing the energy from 2 MeV to 10 MeV a 3.5 % reduction of the $(w_{\text{air}})_p$ value could be expected. A comparable reduction of about 3 % is predicted by the more recent calculations of Grosswendt and Baek (1998).

The ECHED (Vynckier et al 1991, 1994) adopted the value of 35.2 eV from Larson (1958) through ICRU report 31 (1979), a value which they found confirmed in more recent experiments by Hiraoka et al. (1988) and Denis et al. (1990), but also stimulated measurements with calorimeters in order to obtain more information on the $(w_{\text{air}})_p$ -value. Soon after, a number of comparisons of ionisation chamber dosimetry and calorimetry were performed by Siebers et al. (1995), Palmans et al. (1996) and Delacroix et al. (1997). They

all assumed perturbation correction factors for the ionisation chamber in the proton beam to be unity and used the ICRU report 49 stopping power data. The results were all in good agreement with the value recommended in AAPM report 16 as can be seen in table 1. Calorimetry is in principle only able to determine the product of the medium-to-air stopping power ratio, the $(w_{\text{air}})_p$ -value and the perturbation correction factor, but since at that time protocols were under development using the same kind of absorbed dose calorimeters as primary standards, the use of the AAPM report 16 value would result in an improved consistency.

Reference	Energy (MeV)	$(W_{\text{air}})_p$ or $(w_{\text{air}})_p$	1σ
Bakker and Segrè (1951)	340	34.4 ^(*)	0.4
Larson (1958)	1.83	35.2	0.2
Petti et al. (1986)	150	34.2 ^(*)	0.4
Hiraoka et al. (1988)	68.2	35.3	0.7
Denis et al. (1990)	65	35.6	0.6
Siebers et al. (1995)	180	34.4	0.4
Palmans et al. (1996)	55	34.3	0.4
Delacroix et al. (1997)	33, 186	34.3	0.4

Table 1. Experimental values of $(W_{\text{air}})_p$ or $(w_{\text{air}})_p$ for protons in the energy range of clinical proton beams as reproduced in ICRU report 59 (1999). The values are given as reported in the references, except if the stopping power ratios are involved in the determination in which case they are converted to correspond with ICRU report 49 stopping powers. The values marked with a ^(*) are derived from the value measured for argon and the known ratio between argon and air.

Unfortunately, ICRU report 59 decided to adopt a value of 34.8 eV, without giving a solid scientific foundation for it. According to Andreo et al. (2000b) it was obtained as 'an ideal best guess... near the global average of existing data'. The value of 34.23 eV recommended in IAEA TRS-398, on the other hand, is calculated using a rigorous statistical method using weighted medians as explained by Müller (2000). This value offers the additional advantage that it is in good agreement with the calorimetric data and thus improves the inherent consistency of the protocol. Of course it is very unfortunate that these two recent protocols do not propose the same value. As will be clear below, the choice of the $(w_{\text{air}})_p$ -value influences the outcome of certain values such as the relative effectiveness of alanine, which used to confuse the results before the two new recommendations were published. Recent dosimetry studies now tend to choose one of both new values, which only adding to the already existing confusion. More recent values have been measured by Moyers et al. (2000), Palmans et al. (2004a) and Hashemian et al. (2004) confirming the value from IAEA TRS-398 within the experimental uncertainties.

2.2.2.3. Perturbation correction factors

As mentioned before, all CoP's ignore ionisation chamber perturbation correction factors in proton beams, or assume them to equal unity (IAEA TRS-398), which are reasonable assumptions given the ballistic properties of protons. However, some theoretical and experimental studies revealed perturbations of the order of 1%. We have to distinguish

between three sources of perturbation: perturbation of the primary particle fluence, perturbation of the secondary electron fluence and perturbation of the secondary particle fluence from non-elastic nuclear interactions.

Chamber	Effect of		
	<i>cavity only</i>	<i>cavity + wall</i>	<i>cavity + wall + c.e.</i>
Capintec PR06C (Mobit et al. 2000)	2.7	2.5	2.2 2.0 ± 0.4
PTW-30006 and PTW-30001 (Hartmann et al. 1999) (Jäkel et al. 2000) (Kanai et al. 2004)	2.6	2.5	2.3 3 ± 1 2.2 ± 0.2 2.5 ± 0.1
NE2571	2.7	2.5	2.3
NE2581	2.7	2.7	2.0
FWT-IC18	2.0	1.7	1.3
Exr-T2	4.1	3.9	2.7

Table 2. Contributions by the cavity, wall and central electrode (c.e.) to the shift towards the source of the effective water depth of some cylindrical ionisation chambers in mm (columns 2-4) for proton beams. The three references give an experimental value for the chamber type listed on the preceding line.

The first type of perturbations can be considered to consist of gradient effects only, given the low-scatter properties of protons. For cylindrical ionisation chambers, they have been studied extensively by Palmans (1999, 2000) and Palmans and Verhaegen (1998, 2000) using analytical models and Monte Carlo calculations. It was found that gradient effects could be accurately dealt with by taking into account an effective water equivalent depth (which equals the effective point of measurement for a theoretical air cavity surrounded by water). The theoretical results were confirmed in an experiment by Palmans et al. (2001) and by comparison with the experiments of Hartmann et al. (1999) and Kanai et al. (2004) for a PTW-30001 chamber, by Jäkel et al. (2000) for a PTW-30006 chamber and by Mobit et al. (2000) for a Capintec PR06C chamber. All these data are shown in table 2.

Laulainen and Bichsel (1972) and Casnati et al. (1998) studied electron perturbation effects for plane parallel chambers by analytical calculations and Verhaegen and Palmans (2001) devote a Monte Carlo study to it for cylindrical chambers. They all obtained perturbation effects of the order of 1-2% for A-150 walled chambers, although the sign of the perturbations obtained by Casnati et al. (1998) were opposite to those of the other investigations. Recently, Palmans and Verhaegen (2002) proposed an analytical procedure to derive these perturbation correction factors based on a cavity theory model and found it to be in good agreement with the Monte Carlo calculated values. In general, the calculated corrections were never found to exceed 1 % and the relative correction factors even not 0.5 %. The consistency is thus not severely taxed by not using them.

Experiments were performed by Medin et al. (1995) in a 170 MeV proton beam and by Palmans et al. (2001) in a 75 MeV proton beam. They both determined a perturbation correction factor for the presence of an aluminium central electrode in a Farmer-type

geometry. Medin et al. (1995) measured a value of $p_{\text{cel}} = 0.997 \pm 0.004$, whereas Palmans et al. (2001) found a value of $p_{\text{cel}} = 1.000 \pm 0.002$. Palmans et al. (2001) compared the perturbation correction factor of a variety of cylindrical chamber types with the one for a NE2571-type chamber and found in general a good agreement. Also a re-analysis of previously reported data (which were not analysed in this sense in the original publications) resulted in similar conclusions, although the major conclusion was rather qualitative than quantitative: A-150 walled chambers systematically exhibit a lower dose response compared to graphite-walled chambers.

The third type of perturbation correction factors has not been investigated so far. For proton beams, we can assume that they are probably small because most heavy particles produced in non-elastic nuclear interactions have low-energies and thus deposit their energy close to the point where they are generated. The particles that would most likely be responsible for a significant perturbation are secondary protons. Medin and Andreo (1997) found that they are responsible for a change smaller than 0.1% on the water to air stopping power ratio.

2.2.2.4. k_Q -values

The absorbed dose based CoP's need absorbed dose beam quality conversion factors k_Q which essentially convert the calibration factor in the ^{60}Co calibration beam quality into the calibration factor for the clinical light-ion beam quality. ICRU report 59 (1998) gives a procedure to calculate k_Q -values for proton beams although Medin et al. (2000) pointed out that a major error is introduced by neglecting the perturbation in the ^{60}Co calibration beam. TRS-398 lists a detailed table of values for a wide range of cylindrical and plane-parallel chamber types in light-ion beams. Vatnitsky et al. (1996a) were the only to report measured k_Q -data in a proton beam, which were within the uncertainties in good agreement with the theoretical values of IAEA TRS-398.

2.2.2.5. Experimental ionisation chamber dosimetry comparisons

It is interesting to briefly overview a number of dosimetry intercomparisons that has been performed using different ionisation chambers. The maximum chamber-to-chamber variations that result from a number of dosimetry intercomparisons give an idea about the consistence that is in general achieved in proton dosimetry. These results are shown in table 3 for a number of dosimetry studies: Kacperek et al. (1990, 1992), Jones et al. (1992, 1994), Medin et al. (1995), Vatnitsky et al. (1996b), Palmans et al. (1996), Hiraoka et al. (1997), Vatnitsky et al. (1999b), Cuttone et al. (1999), Palmans et al. (2001) and Nohtomi et al. (2001). Comparison of these results is, however, difficult since they are performed at different moments in time applying different dosimetry recommendations and various ionisation chambers, sometimes also in different proton centres. This was also clearly illustrated by Vatnitsky et al. (1996b), who illustrated that if all participants in the intercomparison would have adopted a common protocol, the maximum differences in dose determination would decrease from 6% to 3%. Therefore adoption of a common protocol by all proton centres should be recommended in order to minimize the differences in proton dose determination. On the other hand, an external dosimetry audit must also be recommended for every new proton therapy centre in order to exclude any systematic error in the application of the chosen protocol.

Medin et al. (2000) compared ICRU report 59 and IAEA TRS-398 on a pure theoretical bases and pointed out that in addition to the 1.7 % difference in the $(w_{\text{air}})_p$ -values, a difference of 0.6 % is introduced on the ratio of $(w_{\text{air}})_p$ and $(W_{\text{air}})_c$ due to the (incorrect) use

of values for humid air in ICRU report 59. Combined with the errors quoted before, this gives rise to overall differences between ICRU report 59 and IAEA TRS-398 of -1.5 % to +2.1 % for absorbed dose based dosimetry.

	Maximum difference (%)	Number of chambers	Beam energy (MeV)
Kacperek et al. (1992)	3.5	8	60
Jones et al. (1992)	2.5	7	60-80
Jones et al. (1994)	1.5	8	190
Medin et al. (1995)	0.6-1.6	7	170
Vatnitsky et al. (1996b)	2.8	13	100-250
Palmans et al. (1996)	1.1	10	85
Hiraoka et al. (1997)	1.3	6	70
Vatnitsky et al. (1999)	4.0	20	135-155
Palmans et al. (2001)	1.6	17	75
Nohtomi et al. (2001)	1.5	5	250

Table 3 Maximum chamber-to-chamber absorbed dose differences observed in a number of experimental comparisons involving ionisation chamber dosimetry. Numbers are given for indicative purposes and not to be compared with each other (see text).

The use of plane-parallel chambers in proton beams has been successfully demonstrated by Wu et al. (1995), Medin et al. (1995), Vatnitsky (1999), Mobit et al. (2000), Besserer et al. (2001), Palmans et al. (2002a), Vatnitsky et al. (2002), Kacperek et al. (2002) and Newhauser et al. (2002).

Ionisation chamber measurements in heavy ion beams were reported by Fukumura et al. (1998), Hartmann et al. (1999), Jäkel et al. (2000) and Besserer et al. (2001) and Kanai et al. (2004).

2.3. Primary dosimetry methods

For photon and electron beam dosimetry, mainly three techniques are used as primary dosimetry methods: calorimetry, chemical dosimetry and ionisation chamber dosimetry. For light-ion beams the same methods have been used and in addition methods based on fluence measurements. In this section the application of these four methods to light-ion beam dosimetry is discussed.

2.3.1. Fluence measurements

This method consists in counting the number of light-ion particles N in a field in some way and an accurate determination of the area A (in cm^2) of the beam. From these data, the surface dose in a medium by the light-ion beam could then be calculated as

$$D_{\text{med}} = \frac{N}{A} \cdot (S/\rho)_{\text{med}} \cdot 1.602 \times 10^{-10} \quad (1)$$

where $(S/\rho)_{\text{med}}$ is the stopping power for the medium at the incident particle energy and 1.602×10^{-10} is a factor converting from MeV/g to J/kg. The surface dose can then be related to the dose at the reference point by a (relative) measurement of the depth dose distribution. Note that in equation (1) the stopping power has to include the non-elastic nuclear interaction stopping power, which is not given in ICRU report 49 (1993). An additional difficulty is that, especially when going to higher energies, a secondary particle build-up effect exists over a few millimetres.

The main method used for the fluence determination is using a Faraday cup (FC). Verhey et al. (1979), who compared FC dosimetry with calorimetry and ionisation chamber dosimetry, described a model, which is shown schematically in figure 3 and explains the main features of the device. The diameter of the FC should be large enough, such that the beam is completely captured by the collecting electrode (CE). An electrometer measures the collected charge, from which the number of protons is derived. In order to provide a 100% collection efficiency, the CE should be thick enough, in order to stop all the charged particles that are generated in it. A guard electrode (GE), put on a negative voltage with respect to the CE, prevents backscatter electrons that are generated in the CE from escaping and repels electrons generated in the entrance window and the residual gas in the vacuum between the entrance window and the CE. Windings can generate a magnetic field in the FC, which improves the collecting efficiency by trapping the electrons and forcing them to be deposited close to the generation point. Other comparisons have been described by Verhey et al. (1979), Kacperek et al. (1990), Grusell et al. (1995), Cambria et al. (1997), Delacroix et al. (1997), Jones et al. (1999), Cuttone et al. (1999) and Newhauser et al. (2002), in general yielding an agreement within a couple of percents with other methods, but in some occasions disagreements of 5-10% were demonstrated. For this reason it became less and less recognized as a valuable primary method.

Another method that has been used to determine the fluence is nuclear activation using the $^{11}\text{C}(\text{p,pn})^{12}\text{C}$ reaction. By measuring the activity of ^{11}C in an irradiated ^{12}C slab, the number of incident ions could be derived when knowing the production cross section of ^{11}C . This technique has been applied by Larsson and Sarby (1987) and by Nichiporov (2003). A similar technique was used to determine the dose in tissue by activity measurements of ^{11}C , ^{13}N and ^{15}O after irradiation of tissue with protons.

2.3.2. Absorbed dose calorimetry

Calorimetry consists of measuring the temperature rise ΔT_{med} in a medium absorbing energy from ionisation radiation and converting it into absorbed dose to medium D_{med} by

$$D_{\text{med}} = c_{\text{med}} \cdot \Delta T_{\text{med}} \quad (2)$$

where c_{med} is the specific heat of the medium. It has been recognized as the most direct way to measure absorbed dose. The equation assumes that the energy that is deposited is entirely converted into heat and thus, that no exothermic or endothermic change of state takes place. We could distinguish solid-state calorimetry and water calorimetry since absorbed dose to water is the quantity needed for radiotherapy, water is the preferred medium for calorimetry. Water has the advantage of a low thermal diffusivity such that a temperature profile induced by a radiation field persists long enough to measure it. On the other hand it has the disadvantage of a higher heat capacity than solid materials like graphite and plastics, reducing the sensitivity by the same amount. The (three orders of magnitude) higher thermal conductivity of graphite compared to water compel to using an isolated sample of absorbing material in graphite calorimetry. This way, the average dose over the entire core is determined. In general, correction factors have to be applied to equation 2 for non-medium

materials or vacuum gaps present in the calorimeter, for the non-uniformity of the dose distributions, for conductive and convective heat transfers and for the chemical heat defect if radiation induced chemical reactions take place in the absorption material.

For solid-state calorimetry in light-ion beams, A-150 and graphite have been used as absorbing materials. Fleming and Glass (1969), McDonald and Goodman (1982) and Schulz et al. (1990) used a technique with a double absorber consisting of aluminium and A-150 to determine the chemical heat defect of A-150. They consistently found an endothermic heat defect and derived a correction factor of 1.040 ± 0.015 for an A-150 calorimeter. Schulz et al. (1990) measured the chemical heat defect for graphite using the same technique, confirming confirms within the experimental uncertainty the idea that the chemical heat defect in graphite is zero.

Only one comparison of A-150-calorimetry with ionisation chamber dosimetry in proton beams has been reported. Delacroix et al. (1997) measured a ratio of 0.95 between D_{A-150} measured with the calorimeter and D_{A-150} obtained with eight A-150 walled ionisation chambers. When implementing the more recent stoppers of ICRU report 49 (1993), the ratio was 0.985 ± 0.007 . Earlier, McDonald and Domen (1986) used a similar calorimeter in comparison with ionisation chamber dosimetry in a carbon beam. Measurements were performed at two depths in a mixed water-A-150 phantom. The ratio of the calorimeter to ionisation chamber response reported was 0.98 and 0.99 at the two depths, which is consistent with the results of Delacroix et al. (1997). However, no information on the assessment of the ionometric dose was given.

In the nineties, a number of comparisons of water calorimetry and ionisation chamber dosimetry were performed in proton beams. Schulz et al. (1992) used a water calorimeter in which high purity water was enclosed in a glass phantom and kept at 4° C, this way eliminating any concern related to convective water currents. They reported measurements using this calorimeter in a 160 MeV range-modulated proton beam comparing them with two ionisation chambers. Applying AAPM report 16 for the ionisation chamber measurement, they found a calorimetry to ionisation chamber dose ratio of 0.99 ± 0.01 . If this result is re-analysed with the data in the ECHED CoP (Vynckier et al. 1991, 1994), the ratio becomes 0.98 ± 0.01 . Siebers et al. (1995) used the same calorimeter to determine absorbed dose-to-water in a 250 MeV non-modulated proton beam. From the comparison with two ionisation chambers they derived values for $(w_{air})_p$ between 34.2 ± 0.5 eV and 34.6 ± 0.6 eV depending on the calibration method. On the average, their results was equivalent to a calorimetry to ionisation chamber dose ratio of 0.978 ± 0.005 when applying the ECHED CoP. Seuntjens et al. (1994) and Palmans et al. (1996) used a Domen-type sealed water calorimeter (Domen, 1994) operated at 4° C in a range-modulated 85 MeV proton beam. A considerable amount of original work was done on the chemical heat defect in proton beams, which is discussed below. The calorimetry measurements were compared with ionisation chamber measurements, using three A-150 walled ionisation chambers and applying the ECHED CoP. The calorimetry to ionisation chamber dose ratio was found to be 0.974 ± 0.009 . Another recent comparison between water calorimetry and ionisation chamber dosimetry was reported by Jones et al. (1999). The water calorimeter was of the Schulz-type and ionometry was performed with a Markus-chamber. They found an agreement of 0.4 - 1.8 % in four beam energies around 150 MeV at a depth of 10 cm, but they did not give details on the method that was used for the ionisation chamber measurements. The estimated uncertainty on the comparison was 2 %. Brede et al. (2000) developed a water calorimeter with a very different design, which is also intended to be used in proton beams.

Seuntjens et al. (1994) and Palmans et al. (1996) performed a systematic study of the chemical heat defect for water in proton beams, which had not been done prior to their work. The methods to obtain information on the chemical heat defect were comparable to a study performed by Klassen en Ross (1991). A number of chemical water systems were used in the water calorimeter and the experimental relative dose response was compared with the relative response obtained from numerical simulations. Three chemical systems were investigated: pure water (saturated with argon, which is chemically inert), pure water saturated with hydrogen and a 1mM Sodium Formate (NaCOOH) aqueous solution saturated with oxygen. The influence of the high-LET component in a modulated proton beam on the yields of primary species by ionisation of water was taken into account by calculating the LET as a function of time. As in high-energy photon beams, the results of these calculations showed for the pure water system an initial endothermic heat defect (positive h), which, after a small dose accumulation of less than 10 Gy, evolved to a steady state value of zero. For the hydrogen-saturated system, a small endothermic heat defect was predicted for the first measurement, whereas for all the subsequent measurements it should be zero as well. However, if small initial oxygen concentrations were present, the pure water system showed an initial exothermic behaviour evolving to zero and the hydrogen system showed an initial large exothermic heat defect that could reach 10% or more and which was critically dependent on the initial oxygen concentrations. In both cases a zero heat defect steady state value was reached again. For these reasons it was assumed that when a steady state dose value was reached in the calorimeter measurements, it had reached a zero heat defect. For both systems, the predicted initial behaviour was in agreement with the measurements and both finally reached a steady state dose. The conclusion that both systems reached a zero heat defect was strengthened by the relative agreement of the predicted relative response of the Format/oxygen system with both other systems (the Formate in the system should be regarded as a deliberately added amount of impurities). Brede et al. (1997) designed an experiment to determine the chemical heat defect of water for proton beams in an absolute way. The principle is very similar as the one used to determine the heat defect of A-150 using a combined aluminium/water absorber. A 19 MeV proton beam was completely absorbed either in the water part or in the aluminium part. Preliminary results showed a chemical heat defect that was smaller than 1%.

2.3.3. Chemical dosimetry

Ferrous sulphate used to be a widely used chemical dosimeter for absolute dosimetry, also known as the Fricke dosimeter. An early example of the use of Fricke in heavy ion beams was the measurement of the G value (chemical yield of Fe^{3+} ions per 100 eV deposited energy) in a nitrogen beam by Schimmerling et al. (1976). Besserer et al. (2001) calibrated a ferrous sulphate or Fricke solution in a 14 MeV proton beam against an ionisation chamber and found a value for eG that is in close agreement with a value they extracted from Klassen et al. (1999) for 50 kV x-rays. This result could be understood by the fact that the maximum energy of the secondary electrons is about 30 keV and the electron slowing down spectra could thus be very similar.

Another chemical dosimeter that has already been in use since the 1960's for dosimetry of light and heavy ions, has gained renewed attention as a potential absolute dosimeter is the alanine/EPR system. Mittchel et al. (1966) already reported a good agreement between alanine response and other detectors in high-energy proton beams. More recently, Fattibene et al. (1996) and Onori et al. (1996, 1997) used alanine pellets calibrated in a ^{60}Co beam to determine the dose in the centre of the SOBP of a 60 MeV modulated proton beam (effective energy ± 45 MeV) and in the entrance region of a non-modulated beam (effective

energy ± 60 MeV) and compared it with the response of a Markus-type plane-parallel ionisation chamber. They systematically found a 3% lower dose with their alanine system, which is within the overall uncertainty of the ratio that was estimated to be 6% containing a contribution of 4% on $(w_{\text{air}})_p$. They thus derived a relative effectiveness of alanine of 0.97. However, for the ionisation chamber they used the $(w_{\text{air}})_p$ -value of the ECHED (Vynckier et al. 1991, 1994), which is 3% higher than the current best value resulting from comparison of ionisation chamber dosimetry with calorimetry. They also noticed a slight decrease of the relative effectiveness with decreasing proton energy. Cuttone et al. (1999) performed a similar comparison, using a Markus plane-parallel and an Exradin T1 cylindrical ionisation chamber at effective energy of about 40 MeV and 60 MeV and found an agreement within 2%, but not of a consistent sign. Again this agreement is better than the quoted combined uncertainty of 5% on the ratio of both results. Bartolotta et al. (1999) compared alanine with a Markus chambers at lower effective energies of 24 MeV and 14 MeV and found that the alanine dose-response was respectively 4% and 8% lower. This is another indication that the efficiency decreases already significantly in the BP of a low-energy proton beam. Waligorski et al. (1989) combined experimental data from several publications (Bradshaw et al. 1962, Ebert et al. 1965, Henriksen 1966, Hansen and Olsen 1985 and Hansen et al. 1987) on the variation of the effectiveness with energy for protons and heavy ions and compared these data with results of calculations based on track structure theory. Their results indicate the same qualitative decrease of alanine response with decreasing energy (or increasing LET). For protons the theoretical and experimental results for protons show some discrepancy. The theoretical values are in general higher than the experimental values, but lower than the results of Bartolotta et al. (1999). For higher atomic number ions the agreement between model and theory was in general better.

2.3.4. Extrapolation chamber dosimetry

Zankowski et al. (1998) described the application of an extrapolation chamber as primary dosimetry instrument in a 250 MeV non-modulated proton beam. The agreement with a Farmer type ionisation chamber was within 5%. Another application of a variable volume ionisation chamber was described by Boles et al. (1969), but it was only used for relative dosimetry.

2.3.5. Equivalence of plastic phantoms and water

Measurements of absorbed dose-to-water in plastic phantoms is in general not recommended as it is in the new IAEA TRS-398 Code of Practice. However, in situations where the positioning accuracy is very critical, it could be more convenient to measure in a plastic phantom. This could be the case when performing a measurement on a small SOBP used for ocular therapy or in the presence of a gradient. In principle, dose to water could be derived by scaling the depths in the plastic phantom to a water equivalent depth and by applying the water-to-air stopping power ratio instead of the plastic-to-air stopping power ratio. When the particle fluence is different at the equivalent depths, a fluence correction factor should be applied. Palmans and Verhaegen (1997) indicated that the attenuation of the primary beam due to inelastic nuclear interactions could be different in both materials and thus could contribute to differences in fluence. Palmans and Vynckier (2000) and Palmans et al. (2002b) performed Monte Carlo calculations to confirm this and compared the results with measurements. Their results suggest that a measurement in a low-energy clinical proton beam could be done in a plastic phantom without a substantial loss in accuracy. For higher energy beams, the fluence corrections could be substantial and amount to 3-5 % for a very

deeply penetrating beam, depending on the data set. Recent experimental results by Moyers et al. (2002) and Schneider et al. (2002) also confirm these results.

2.4. Relative dosimetry

Besides an accurate determination of the absolute dose at a reference point, a radiotherapy treatment planning requires also the spatial distribution of the dose deposition. This necessitates the measurement of depth dose distributions, lateral dose distributions and output factors (ratio of the dose measured at the centre of a field compared to the dose measured in the centre of a reference field). These measurements should ideally all be carried out in water, but as for the absolute measurements it could be advisable to do measurements in a plastic phantom if extremely accurate positioning is required. This is in general the case in low-energy beams, such as the proton beams for ocular therapy. Given the low scatter properties of light-ions, output factors will in general be unity except for very small fields such as used in proton radiosurgery.

The ideal detector has not been found and will on the one hand depend on the particle type, its energy and the extension of the field and on the other hand on the geometrical properties and the energy dependence of the detector. In general, if possible, an ionisation chamber should be used (IAEA TRS-398). For depth dose distributions a plane-parallel chamber is preferred and if a cylindrical chamber is used an effective point of measurement should be taken into account. The ionisation as a function of depth should also be corrected for the variation of the water to air stopping power ratios with depth. If another detector is used, the dependence of its response with energy should be checked against an ionisation chamber.

Vatnitsky et al. (1999a) compared several detectors for dosimetry of narrow proton beams and their results could be used as a guideline. They recommend radiochromic or radiographic film for the measurement of lateral dose distributions. For depth dose distributions small volume ionisation chambers can be used down to beam diameters of 20 mm, thermoluminescent dosimeters (TLD) and diamond detectors could be used down to 10 mm and diodes down to 5 mm. For smaller field sizes only film can be used.

The following subsections give an overview of various detectors that have been evaluated for relative dosimetry and their advantages and disadvantages.

2.4.1. Alanine

Its absolute response and energy dependence have been discussed in section 2.3.3. Several authors have investigated its suitability for depth dose measurements compared to other detectors (Nichiporov et al. 1995, Gall et al. 1996, Fattibene et al. 1996, Onori et al. 1996, 1997, Bartolotta et al. 1999) in general with satisfying agreement with plane-parallel ionisation chambers. The advantage of this detector is that the stopping power over the entire clinical proton energy range is close to that of water (Onori et al. 1997) and whilst the density of pure alanine is quite high (1.42 g cm^{-3}) compared to water, the density of the mixtures in which it is in general used for practical purposes is much closer to water. The pellets in which it is often compressed together with paraffin wax, polyethylene or cellulose as a binding material have typical densities varying from 1.05 to 1.26 g cm^{-3} . Alanine-film, in which typically 30% by weight alanine grains are bound in polyethylene have relative densities around unity. Several authors confirmed the linearity of alanine pellets in the accumulated dose range of 5 to 300 Gy (Nichiporov et al. 1995, Gall et al. 1996, Onori et al. 1997). A disadvantage is that the sensitivity of the technique is not particularly high. Pellets of 2 mm thickness and 5 mm diameter require doses of more than 10 Gy for a standard

deviation on the read-out of 1%, whereas alanine films require over 100 Gy for reducing the standard deviation below 10% (Onori et al. 1997). For pellets, another general disadvantage observed and admitted by most of the quoted authors is the volume averaging effect in low-energy beams due to the relatively large size. When measuring a depth dose curve using a stack of pellets in a solid phantom (PMMA or solid water), Onori et al. (1997) also observe dose tails beyond the BP, which they explain as due to air gaps between the detectors and the surrounding material. This could partially contribute to the effect, but it can be shown that even with perfectly matching detectors in-scatter effects can explain the major contributions to the tails (see also the discussion on TLD below). Furthermore, in none of the investigations an attempt was made to assess the perturbation factors for alanine pellets.

2.4.2. Thermoluminescent dosimeters

TLD have been widely used in light-ion beams for the measurement of depth dose curves since decades (Olko et al. 2004). In general the detectors consist of a crystalline material such as LiF, CaF₂, CaSO₄ and CsI doped with various elements such as P, Mg, Ti, Cu, Tm,... In general they are produced in small pellets or chips, some of which have been specifically designed for measurements in the BP region of proton beams by making them very thin, down to 0.1 mm (Bilski et al. 1999).

Some early experiments for depth dose curve measurements were reported by Carlsson and Carlsson (1970) and Momeni et al. (1973). After that, several authors investigated various parameters such as the response versus accumulated dose and the energy dependence.

Track structure theory predicts that TLD show supralinearity. Horowitz (1990) performed a systematic theoretical analysis of this effect. Bucciolini et al. (1999, 2000) measured the dose-response for accumulated doses below 20 Gy. Benabdesselam et al. (2000) investigated the dose-response from 10 to 100.000 Gy for alphas and from 100 to 100.000 Gy for protons both TLD-100 and TLD-500.

In general the efficiency of TLD detectors decrease with increasing LET. Carlsson and Carlsson (1970) compared depth dose curves measured with LiF-TLD and an ionisation chamber and Fricke to derive the sensitivity. They also investigated the influence of the variation of stopping power with energy. Horowitz and Stern (1990) and Avila et al. (1999) studied the LET dependence of the TLD response with track structure theory, in a similar way as Waligorski et al. (1999) did for alanine. They both found a good agreement between experiment and theory for both protons and alpha particles, although the theoretical LET dependence was stronger than found in the experiments. Spurny (1992) measured the efficiency for various TLD materials in protons of energies larger than 50 MeV and found no significant energy dependence in that region. Vatnitsky et al. (1995) determined the relative sensitivity of a diamond detector in a 250 MeV proton beam in the centre of an SOBP and in the BP and found it to be unity within 2%. Bilski et al. (1997) determined experimental relative efficiencies for various TLD materials in an alpha particle beam. Besserer et al. (2001) tabulated light attenuation correction factors and energy dependence of TLD-100 (MTS-N) as a function of proton energy. They also evaluated TLD as reference dosimeter, which was reasonably successful although the uncertainty of the comparison was about 8%. Some authors investigated the different energy dependence of the different glow curve peaks (Schmidt et al. 1990, Hoffmann 1996, Loncol et al. 1996a, Loncol et al. 1996b and Hoffmann et al. 1999).

The effect of density scaling was investigated by Fattibene et al. (1996) and Torrisi et al. (1997a). Fattibene et al. (1996) proposed scaling with the electron densities. The also showed the dose tails beyond the BP due to in-scattered protons from the surrounding

medium. Palmans et al. (1997) showed that it is more appropriate to scale with the continuous slowing down ranges and that the dose tail can be predicted and understood by Monte Carlo calculations.

2.4.3. Diodes and diamonds

Koehler (1967) was the first to use silicon diodes for dosimetry of proton beams. Maruhashi and Nakamura (1980) developed a miniature p-n junction silicon diode for which they determined the sensitive volume in a proton beam and used it to measure radial distributions of proton beams in several low-Z materials and iron. They also used it to measure a 2-D dose map in water. Various characteristics of MOSFET in proton beams were investigated by Litovchenko et al. (1990). Grusell and Medin (2000) studied various silicon diodes for depth dose curve measurements and found that only Hi-p type diodes show a response proportional to the ionisation in the silicon crystal. They also calculated stopping power corrections for this type of diode.

Khrunov et al. (1990) were the first to apply diamond detectors in proton beams. Only a lateral dose measurement is shown. The diamond measurements exhibit a sharper penumbra, probably due to the sensitive volume. Vatnitsky et al. (1993, 1995) determined the relative sensitivity of a diamond detector in a 250 MeV proton beam and found it to be unity within 2%. Onori et al. (2000) investigated the linearity, reproducibility, dose rate dependence, energy/LET dependence of a diamond and a very thin p-type silicon diode in low-energy proton beams (8.3 to 21.5 MeV). They found that mainly the energy dependence is an issue and that it is very comparable for both detectors. The diamond, however, substantially distorted the depth dose distributions due to its size (0.26 mm!). Pacilo et al. (2002) performed a similar study in the 10 to 59 MeV range but found a larger sensitivity change for diamond compared to diode at low proton energies. Besides that, both detectors were of even value. Fidanzio et al. (2002) studied the dose rate dependence of a diamond detector and found that it is not LET-dependent, such that values measured for photon beams could be used for protons. They also investigated the energy dependence for a non-modulated proton beam and showed that corrections could be derived for a depth dose curve measurement of a modulated proton beam by integrating the correction factors over the particle spectrum. A literature survey of experimental data showed that these correction can amount to 20% for the diamond detectors and diodes.

2.4.4. Radiographic and radiochromic film

Daftari et al. (1999) tested GafChromic MD-55 film for lateral and depth dose measurements. The results were not very good, mainly because they were not corrected for inhomogeneities of the film response, a well known problem with this type of films. They reported a significant build-up of optical density after irradiation that lasted for more than 100 hours. Vatnitsky et al. (1997) reported similar behaviour for the same type of film.

Hansen and Olsen (1984) measured and calculated the relative efficiency (RE) of the radiochromic dye film in the same way as they did for alanine. They found significant deviations from unity for LET higher than $30 \text{ MeV cm}^2 \text{ g}^{-1}$ and a reasonable agreement between theory and experiment. A more detailed analysis by Olsen and Hansen (1984) revealed that the agreement in ^7Li and ^{16}O ion beams is good (better than 20%) for energies above 1.5 MeV/A.U. and considerably worse for lower energies.

Hansen et al. (1984) investigated the time dependence of the optical density after irradiation. The signal build-up lasted more than 10 hours and was dependent on the medium (gas) in

which the irradiation was performed. They also measured the accumulated dose dependence of the film, which was found to be well linear on a log-log scale up to 1000 Gy.

Jones et al. (1999) measured the accumulated dose dependence of GafChromic film, which was found to be well linear on a log-log scale up to 1000 Gy. The film was also used to determine the effective area of the beam, an essential parameter for a Faraday cup measurement.

Piermattei et al. (2000) investigated the use of radiochromic film for the measurement of depth doses in a 21.5 MeV beam. Several orientations were investigated. A measurement with a number of films positioned perpendicularly to the beam at different depths was shown to give a better result than with a film positioned along the beam axis and with a film positioned slightly tilted with respect to the beam axis. This was mainly ascribed to geometrical effects. The considerable difference (30% in the BP) between the first geometry and the ionisation chamber were subscribed to the LET-dependence of the film response.

Spielberger et al. (2001) performed an experimental study of the dose response curve and the LET-dependence for radiographic film in protons and heavy ions. They found a strong dependence on the particle type. The variation with LET they reported is more or less similar as the dependence for radiochromic film reported by others. Spielberger et al. (2002) extended this work and showed that their model could predict the distortion of the BP when measured with radiographic film very well.

Vatnitsky (1997) investigated dose response of radiochromic film up to 100 Gy and measured lateral and depth dose profiles. Consistent with other investigations, a decreasing sensitivity in the BP region or the distal region of an SOBP was observed. Vatnitsky et al. (1997) demonstrated the usefulness of radiochromic film for lateral beam profile measurements.

2.4.5. Gel dosimetry

Johansson et al. (1997) used ferrous sulphate gel for measuring 3-D dose distributions in a clinical radiation. Bäck et al. (1999) investigated ferrous sulphate (Fricke) gel dosimetry for the measurements of dose distributions in a 90 MeV proton beam. In the BP region or the distal region of an SOBP the gel shows a decreasing sensitivity, which can be explained by the LET dependence of Fricke solution. DiCapua et al. (1997) showed that this LET-dependence depends on the matrix (agarose) concentration and even disappeared at a 1% agarose concentration. DiCapua et al. (1997) and Gambarini et al. (1997a, 1997b) also reported an increased yield of Fe^{3+} in protons, which was, however, denied by Bäck et al. (1999).

The first application of a polymer gel (BANG) in light-ions thus far was reported by Ramm et al (2000). They measured depth doses in a carbon beam and observed that also for this type of gel, the efficiency reduces with increasing LET. More recently, Heufelder et al. (2003) investigated the use of BANG polymer gels in a low-energy proton beam and Gustavsson et al. (2004) studied the LET response of normoxic polymer gels (that can operate in the presence of oxygen) to protons.

2.4.6. Other detectors

Various other detectors have been used for relative dosimetry. Boon et al. (1998) used a scintillating screen observed with a CCD camera for 2-D dosimetry. Track etching detectors have been used by Dajkó (1996), silver-activated phosphate glass rods by Mitchell et al.

(1966), imaging plates by Nohtomi et al. (1999, 2000) and plastic scintillators by Torrisi et al. (1997b), Torrisi (2000) and Safai et al. (2004). Cirio et al. (2004) developed a pixel ionisation chamber for two-dimensional dosimetry.

3. EXPERIMENTAL WORK IN THE SCOPE OF THE LIGHT ION DOSIMETRY PROJECT

3.1. Graphite calorimetry

This section combines results published in Palmans et al. (2003, 2004a) and results presented at the June 2004 PTCOG meeting in Paris (Palmans et al. 2004b). The paper Palmans et al. (2004a) is reproduced in addendum A and a brief summary is given here. The results of Palmans et al. (2004b) are added below since they discuss the impact the work has on the present day knowledge of the average energy required to produce an ion pair.

Calorimetry is the preferred method for dosimetry of clinical proton beams as recommended in various reports (Lyman et al. 1986, Vynckier et al. 1991, 1994, ICRU report 59, 1998). A recent code of practice, IAEA TRS-398 (Andreo et al. 2000), recommends dosimetry using an ionisation chamber (IC) with an absorbed dose to water calibration factor in the clinical proton beam or, equivalently, to use a measured beam quality correction factor k_Q . Despite this, no calorimetry based long-term primary standards for protons exist at present.

The feasibility of calorimetry for dosimetry of clinical high-energy proton beams as well as for relatively low-energy beams has been demonstrated. Various calorimeter experiments were performed to evaluate the chemical heat defect in water, graphite and A-150 tissue equivalent plastic (see Palmans and Vynckier, 2002, for an overview). Comparisons of IC dosimetry with water calorimeters (Schulz et al. 1992, Siebers et al. 1995, Palmans et al. 1996) and an A-150 calorimeter (Delacroix et al. 1997) have yielded valuable information on the average energy required to produce an ion pair in dry air by protons. For low-energy proton beams used for the treatment of ocular melanoma, on the other hand, the feasibility of calorimetry has not been explored mainly because of the short range of these beams (less than 3.5 cm) and their limited lateral extension (a few cm in diameter).

In this work, we want to explore the possibility of using a graphite calorimeter for low-energy proton beams. The construction of a small portable graphite calorimeter (SPGC), an adaptation of NPL's existing portable graphite calorimeter (PGC) to the specific situation of a small low-energy proton beam, will be described. The correction factors needed for comparing the SPGC with ICs are investigated and a comparison will be discussed.

3.1.1. Methods: proton beam line, portable graphite calorimeter, ionisation chambers

The 62 MeV proton beam at Clatterbridge Centre of Oncology (CCO) is generated by a cyclotron and used for the treatment of ocular melanoma. A detailed diagram of the beam is given in figure 5 (Bonnett et al. 1993). A description of the components and the beam characteristics is given in addendum A.

The small-body portable graphite calorimeter (SPGC) used in this work is derived from NPL's previously existing portable graphite calorimeter (PGC), which is described in detail by McEwen and Duane (2000) and McEwen (2002). An SPGC was built such that the entire jacket was exposed to the beam. A schematic diagram of the SPGC is shown in figures 6 and 7 and pictures of the set-up in figure 8. The details of the construction and set-up and the analysis of the measurements are discussed in addendum A.

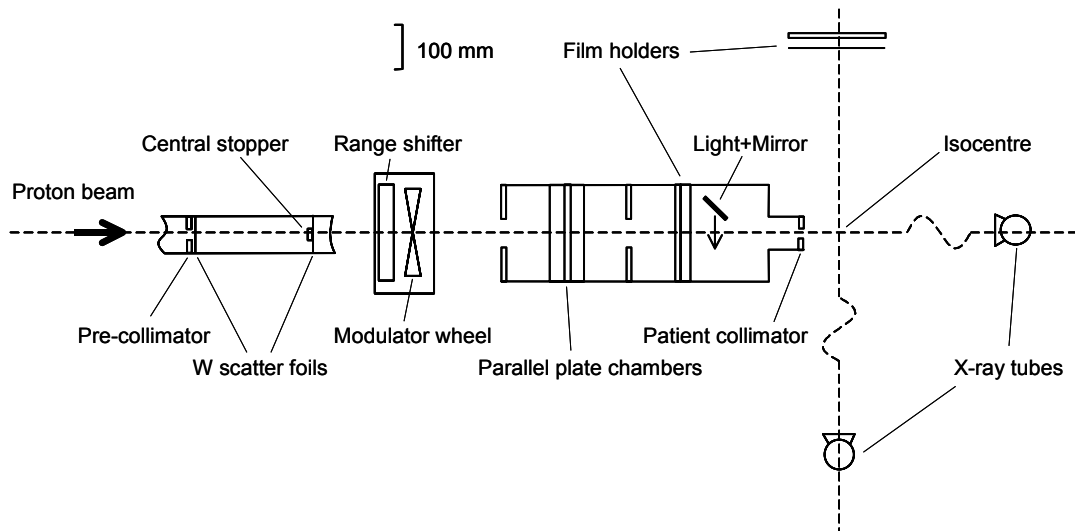


Figure 5. Schematic drawing of the beam line at CCO [Adapted from Bonnet et al., 1993]

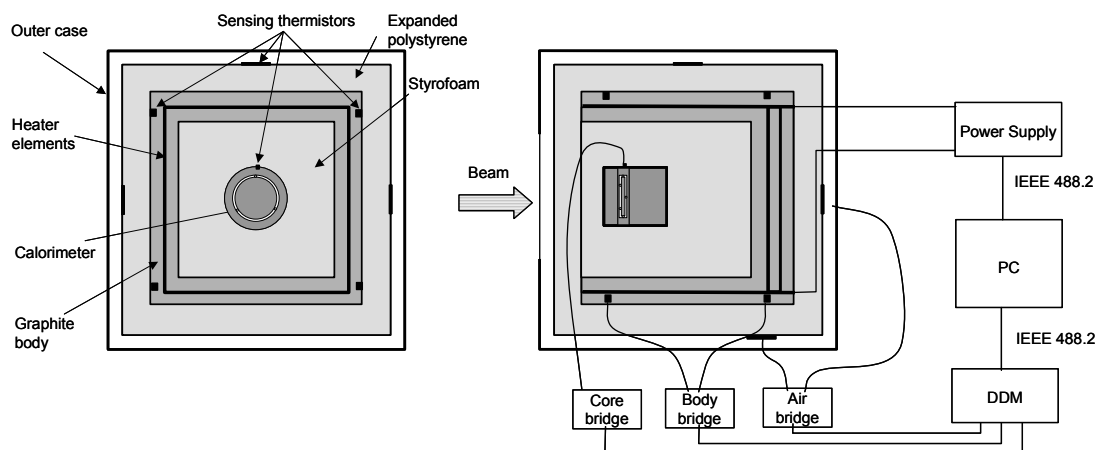


Figure 6. Schematic diagram of the small-body portable graphite calorimeter (not to scale, adapted from McEwen and Duane, 2000) and the temperature control electronics.

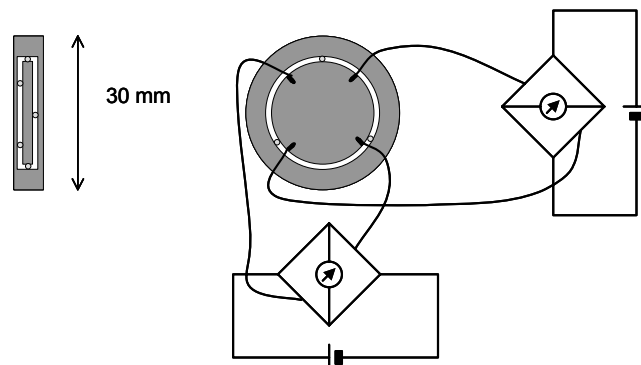


Figure 7. Cross-sections of the core and the jacket (approximately to scale) along the beam axis (left) and perpendicular to the beam axis (right) and schematic representation of the thermistor connections into the Wheatstone bridges.



Figure 8. Left: calorimeter body, with part of its styrofoam surround. The final collimator of the proton beam line is visible at the left. Right: calorimeter housing (blue) in front of the proton beam line, with the team that performed the first ever therapy level absolute dose measurements using a graphite calorimeter in a clinical proton beam; from l. to r.: Mary Simon, Hugo Palmans, Russell Thomas (NPL) and Andrzej Kacperek (CCO)

The calorimeter response was compared with two cylindrical and two plane-parallel ionisation chambers (ICs) with an absorbed dose to water calibration factor in ^{60}Co . The plane parallel ICs had in addition an absorbed dose to water calibration factor in a 19 MeV electron beam. Dosimetry was performed using k_Q -data from the IAEA code of practice TRS-398 (Andreo et al. 2000). The IC measurements were performed in a separate graphite phantom. The details of the set-up and the calculation of various correction factors for profile non-homogeneity, difference in the effective source to detector distance (SDD) and difference in the water equivalent depth of the IC compared to the SPGC are discussed in addendum A.

3.1.2. Results

3.1.2.1. Comparison of the portable graphite calorimeter response with ionisation chambers

Figure 9 shows $D_{w,SPGC}/D_{w,IC}$, the ratio of absorbed dose to water obtained from the SPGC measurements and from the IC measurements after all correction factors discussed above were applied. Each data point represents a set of measurements, which usually consisted of eight calorimeter runs with IC measurements prior to and after the set of calorimeter measurements. The standard deviation on a set of SPGC measurements was typically 0.6%. The uncertainty bars represent combined type A standard uncertainties for the calorimeter and IC measurements. The horizontal dashed lines represent the weighted mean for each IC type, beam quality and calibration modality.

No significant difference was observed between the measurements performed in June 2003 and those performed in November 2003. The different IC types were found to agree well within one calibration modality and beam type. There was, however, a systematic difference between the data obtained with ^{60}Co calibration factors and those obtained with electron calibration factors. This confirms earlier results reported by Thomas et al. (2003) and indicates that the ratio of the calibration factors and the ratio of the IC data from IAEA TRS-398 for ^{60}Co and electron beams are inconsistent. The issue was also raised by Andreo et al. (2002) and is presently under further investigation at NPL.

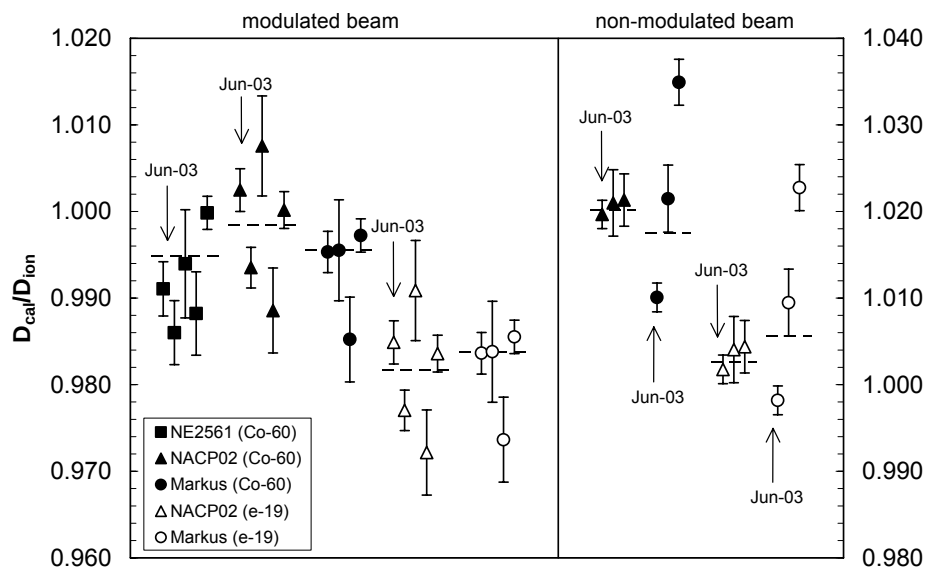


Figure 9. Ratio of dose to water obtained with the SPGC and dose to water obtained with the ICs applying IAEA TRS-398. The data points measured in June 2003 are indicated.

Calibration beam quality	$D_{w,SPGC}/D_{w,IC}$		$(w_{air})_p$ in J/C	
	modulated	non-modulated	modulated	non-modulated
^{60}Co	0.996	1.019	34.1	34.9
e_{19}	0.983	1.004	33.6	34.4

Table 4. Ratio between the dose obtained from the SPGC, $D_{w,SPGC}$, and from the ICs, $D_{w,IC}$ and the $(w_{air})_p$ value derived from these data, accepting the stopping powers and the omission of proton beam related IC perturbation factors in IAEA TRS-398.

For the modulated beam, the average $D_{w,SPGC}/D_{w,IC}$ ratio was 0.996 using ^{60}Co calibration factors and 0.983 using electron calibration factors. For the non-modulated beam it was 1.019 using ^{60}Co calibration factors and 1.004 using electron calibration factors. By accepting the data related to the calibration beams we are in principle only able to derive values for the product of the average energy required to produce an ion pair in dry air by protons, $(w_{air})_p$, the water to air stopping power ratio in the proton beam, $(s_{w,air})_p$ and the overall IC perturbation factor in the proton beam, p_p . Based on the $(w_{air})_p$ value of 34.23 J/C recommended in IAEA TRS-398, we could derive $(w_{air})_p$ by accepting all the other data in IAEA TRS-398 to find values of 34.1 J/C using ^{60}Co calibration factors and 33.6 J/C using electron calibration factors in the modulated beam and 34.9 J/C using ^{60}Co calibration factors and 34.4 J/C using electron calibration factors in the non-modulated beam. The uncertainties on these values are discussed in the next section. These data are summarised in table 4.

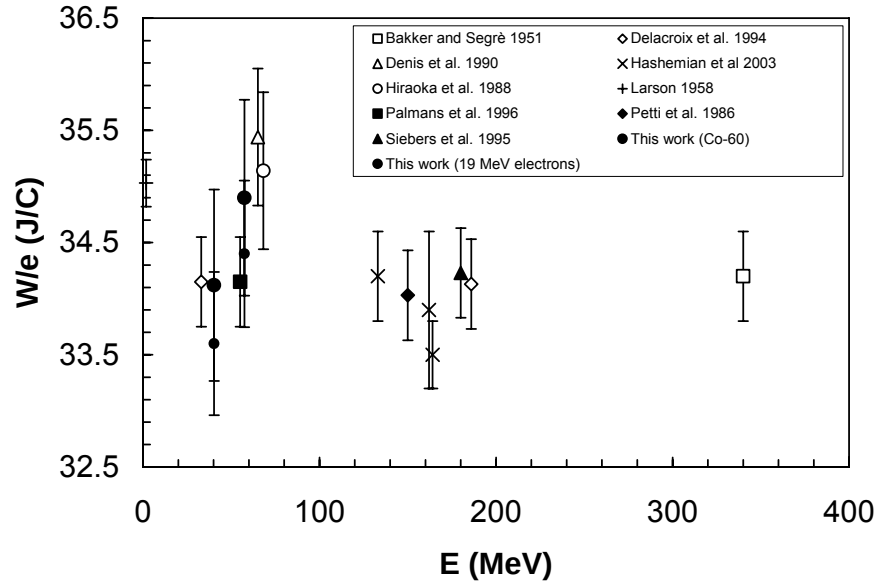


Figure 10. $(w_{\text{air}})_p$ values used in IAEA TRS-398 combined with the values reported by Hashemian et al., 2003 ($\times$ -symbols) and the values obtained in this work.

Figure 10 shows the $(w_{\text{air}})_p$ values obtained in the work, plotted together with the $(w_{\text{air}})_p$ values which were used in IAEA TRS-398 to determine a median value and the values reported at the previous PTCOG meeting by Hashemian et al. (2003). Deriving a $(w_{\text{air}})_p$ from all the data represented in figure does not result in a significantly different value compared to IAEA TRS-398 when using the same procedure of weighted medians.

3.1.2.2. Uncertainties

Table 5 shows a typical uncertainty budget for one of the ICs in a modulated beam. The uncertainties are similar for other IC types and for the non-modulated beam. The major contributions are discussed below.

On the SPGC side, the uncertainty is dominated by the uncertainty on $s_{w,\text{air}}^{\Delta}/s_{\text{gr},\text{air}}^{\Delta}$. A standard uncertainty of 1% was taken from IAEA TRS-398 for proton stopping power ratios in the clinical energy range, which is consistent with the lower values quoted in ICRU report 49. For the ICs, the uncertainty on both the ^{60}Co as the electron beam calibration factors is 0.75%. The standard uncertainties on the k_Q values are 1.7% for the cylindrical IC and 2.1% for the plane parallel ICs. The standard uncertainty on the $k_{Q,Q_{e19}}$ values for the plane-parallel ICs was obtained by combining the uncertainties for the proton beam data and the electron beam data in IAEA TRS-398 and was found to be 1.4%. Other substantial contributions were 0.2% for the reproducibility of the PGC - IC response ratio (0.4% for the Markus in the non-modulated beam) and 0.3% for the lateral profile correction factors (0.5% for the Markus in the non-modulated beam). This results in standard uncertainties on the ratio $D_{w,\text{SPGC}}/D_{w,\text{IC}}$ of 2.1% for the NE2561 and 2.5% for the plane-parallel ICs using ^{60}Co calibration factors, and 1.9% for the plane-parallel ICs using electron calibration factors. These numbers are also the standard uncertainties with which we are able to derive $(w_{\text{air}})_p$. Leaving out the uncertainties on the beam quality correction factors yields 1.3% to 1.4%,

which is thus the standard uncertainty with which we were able to determine k_Q or $k_{Q,Q_{e19}}$ values.

Source of uncertainty	Based on ^{60}Co	Based on Q_{e19}	type
<i>for SPGC</i>			
thermistor calibration	0.10		B
specific heat	0.08		B
$s_{w,g}$	1.00		B
k_{gap}	0.05		A
k_{volume}	0.05		A
extrapolation	0.06		B
<i>for NACP-02</i>			
reproducibility	0.29	0.27	A
positioning	0.30	0.30	B
p_{ion}	0.10	0.10	B
p_{pT}	0.05	0.05	B
k_{SDD}	0.06	0.06	B
k_{pdd}	0.00	0.00	B
k_{fluence}	0.05	0.05	B
k_{profile}	0.21	0.21	B
$N_{D,w}$	0.75	0.75	B
k_Q	2.10	1.40	B
$u(D_{w,\text{SPGC}}/D_{w,\text{IC}})$	2.5	1.9	
$u(k_Q)$	1.4	1.3	
$u(w_{\text{air}})$	2.5	1.9	

Table 5. Uncertainties in % for the comparison of the SPGC with the NACP02 chambers in the modulated beam. Overall uncertainties on the $D_{w,\text{SPGC}}/D_{w,\text{IC}}$ are shown, as well as uncertainties on the experimental k_Q values that could in principle be derived and on the derived $(w_{\text{air}})_p$ values.

3.1.2.3. Recommendations for improvement of the calorimetric standard

The following improvements for the PGC and the measurement set-up in general will enable a substantial reduction of the overall uncertainty:

- reduction of the size of the temperature controlled body will establish quicker settling and will make measurements more efficient by allowing shorter time intervals; a smaller calorimeter will also allow the use of an external monitor between the final collimator and the SPGC or IC phantom in the CCO beam
- IC measurements should be performed in a phantom that allows accurate positioning at the same depth as in the SPGC (especially for non-modulated beams),
- making the core size to match the IC size will allow minimising the uncertainty due to lateral beam non-uniformity,

- removing the copper positioning cross wires, which is the main source of the lateral profile non-uniformity, for precision dosimetry work
- calibrations of the ICs in terms of absorbed dose to graphite directly to the SPGC and a detailed analysis of the uncertainties on all remaining data needed could reduce the uncertainty on the determination of $(w_{\text{air}})_p$.

3.2. Alanine dosimetry:

A literature review was published by Palmans (2003) and is reproduced in addendum B. Work on interaction cross section data was done in support of the interpretation of experimental results and is described below as well as the results of experiments that were performed.

3.2.1. Interaction cross section data

3.2.1.1. Stopping powers

The most comprehensive set of stopping power data available at present are those from ICRU report 49 (1993), which presents, however, no data for alanine. There have been some discussion on the rationale of the selection of certain data between Ziegler (1999) and Seltzer et al. (2001) but this discussion does not affect the low-Z material data. Onori et al. (1997) calculated stopping powers for alanine by applying Bragg's rule to the stopping powers of the elementary constituents. According to this rule, the mass collision stopping power (S_{col}/ρ) for a compound medium could, with a reasonable accuracy, be calculated as:

$$\left(\frac{S_{\text{col}}}{\rho} \right) = \sum_j w_j \cdot \left(\frac{S_{\text{col}}}{\rho} \right)_j \quad (7)$$

where w_j is the fraction by weight and $(S_{\text{col}}/\rho)_j$ the mass collision stopping power of the j 'th constituent (ICRU report 49). Correspondingly, the mean excitation energy I obtained is given by:

$$\ln I = \frac{\sum_j w_j \cdot \left(\frac{Z_j}{A_j} \right) \cdot \ln I_j}{\left\langle \frac{Z}{A} \right\rangle} \quad (8)$$

where

$$\left\langle \frac{Z}{A} \right\rangle = \sum_j w_j \cdot \left(\frac{Z_j}{A_j} \right) \quad (9)$$

The value of I for alanine obtained with Bragg's rule is 69.3 eV. However, this rule does not account for the influence of chemical binding effects on the mean excitation energy. The mean excitation energies used in ICRU report 49 are entirely based on the extensive study of the subject for the electron and positron stopping powers of ICRU report 37 (1984). This

report recommends the use of experimental I -values when a sufficiently accurate value is available. If an experimental value is not available, a calculation using an adapted Bragg rule should be done, in which different I -values are assigned to the constituent elements depending on the physical phase of the material. This way, the mean excitation energy calculated for alanine is 71.9 eV. An alternative presented in ICRU report 37 uses Thompson's assignment scheme that accounts for the type of chemical bond each element exhibits in condensed organic compounds. However, due to the large uncertainties on the elemental data, it does not provide more accurate values than using the adapted Bragg rule applied above. Thompson's rule yields an I -value for alanine of 73.0 eV. There is another argument to be consistent with the mean excitation energies of ICRU report 37: Medin and Andreo (1992) pointed out that, since calibrations of ionization chambers and other detectors are often performed in ^{60}Co gamma beams (thus involving electron stopping power data), it improves the internal consistency of dosimetry when for proton dosimetry the same I -values as for electrons are used.

The method used here to account for the difference is to calculate the stopping power for alanine using Bragg's rule, subtract the term containing the Bragg calculated mean excitation energy I_{Bragg} and add the term with the correct value of I , such that the correction term Δ_I to be applied is:

$$\Delta_I = 0.307075 \cdot \left\langle \frac{Z}{A} \right\rangle \cdot \frac{1}{\beta^2} \cdot \ln \left(\frac{I_{\text{Bragg}}}{I} \right) \quad (10)$$

As will be shown below, this term is the major influence in the clinical energy region from 1 MeV up to 250 MeV.

At higher energies, it is also important to get the density effect right. For that reason, the density effect has been subtracted from the elemental data, prior to applying Bragg's rule, and the density effect for alanine has been calculated separately according to Sternheimer's model outlined in ICRU report 37. To this end, the calculation algorithm from ESTAR was used, which is justified since only material properties (electron levels) and the relative velocity β of the incident particles are involved.

At lower energies, Bragg's additivity rule also induces consistency errors on the calculation of the Barkas correction. This is solely due to the different values for the scaled minimum-impact parameter b that is used for molecular hydrogen and hydrogen in compounds. Therefore an additional correction term has been calculated for this error:

$$\Delta_{\text{Barkas}} = 0.307075 \cdot \left\langle \frac{Z}{A} \right\rangle \cdot \frac{1}{\beta^2} \cdot w_H \cdot \left[\frac{\gamma \cdot Z \cdot \alpha^3}{\beta^3} \cdot \left(F_{\text{ARB}} \left(\frac{b_{\text{comp}} \cdot Z^{1/2} \cdot \alpha}{\beta} \right) - F_{\text{ARB}} \left(\frac{b_{\text{molec}} \cdot Z^{1/2} \cdot \alpha}{\beta} \right) \right) \right] \quad (11)$$

where w_H is the fraction by weight of hydrogen in the compound, γ is a constant ($= 1.29$), α is the fine structure constant ($= 137.036$), b_{comp} ($= 1.8$) and b_{molec} ($= 0.6$) are the values of the scaled minimum-impact parameter for hydrogen in compounds and for molecular hydrogen respectively and F_{ARB} is a function, which was taken from Ashley et al. (1972).

The previous rules were applied to proton energies T above 500 keV. In the low energy region (T lower than 200 keV), which is not really important for macroscopic clinical dosimetry, the data were obtained by applying the bare Bragg additivity rule, as was done for most other compounds in ICRU report 49. The gap region between 200 keV and 500 keV is then filled up by a cubic spline fit through all the data on a Fano plot (i.e. $\beta^2 \times$ stopping power versus $\log T$) as was done in ICRU report 49.

In order to evaluate if these procedures work properly, they were first verified for PMMA, for which data are tabulated in ICRU report 49. The mean excitation energy for PMMA calculated with Bragg's rule is 68.6 eV, whereas the value used in ICRU report 49 is 74.0 eV, being an experimental one. It is obvious that the difference is more pronounced than for alanine. Figure 1 shows the ratio of the calculated stopping powers and the tabulated stopping powers for PMMA, (i) calculated from the elemental data with Bragg's rule only, (ii) with additionally the density correction taken into account properly and (iii) with additionally to the former two, the correction for the I -value according to the equation (11) taken into account.

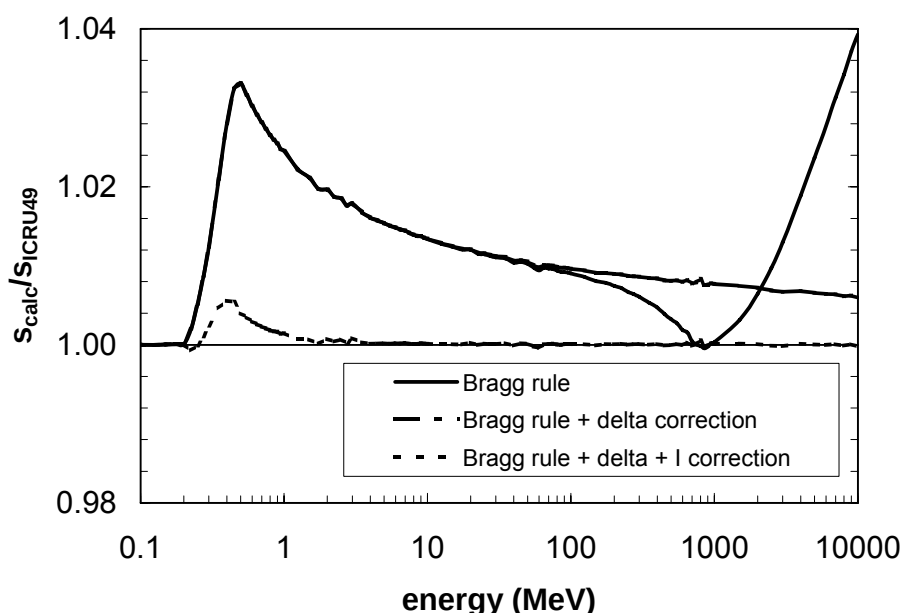


Figure 11. Ratio of calculated and tabulated (ICRU report 49) stopping powers for PMMA as a function of energy

After these two corrections, the agreement of the calculated data and the tabulated data is excellent, except for a bump of 0.6% around 1 MeV. This is entirely explained by the error made on the Barkas correction. After applying equation (11), the calculated and tabulated stopping powers agree within 0.1% over the entire energy range. The remaining differences are probably due to round-off errors in the tabulated data and have no influence on the outcome of a Monte Carlo calculation. We performed a Monte Carlo calculation of depth dose distributions for both the calculated and tabulated PMMA stopping powers and found that the differences induced by the bump are not significant; they are most dominant in the distal edge of the Bragg peak and always smaller than 0.01% of the peak dose. On the other hand, it is clear from figure 11 that the density correction and the I -value play an important role on the stopping power values.

Material	Composition (fractions by weight) or chemical formula	density (g cm ⁻³)	I_B (eV)	I (eV)
alanine	CH ₃ CH(NH ₂)COOH	1.42	69.3	71.9
alanine pellet (Onori et al. 1997)	0.8 × alanine + 0.2 × paraffin wax	1.26	66.3	68.2
alanine pellet (NPL)	0.9 × alanine + 0.1 × paraffin wax	1.15	67.8	70.1
alanine film (Onori et al. 1997)	0.3 × alanine + 0.7 × polyethylene	0.99	59.9	60.6

Table 9 Properties of the different alanine and alanine mixtures that are used in this work .

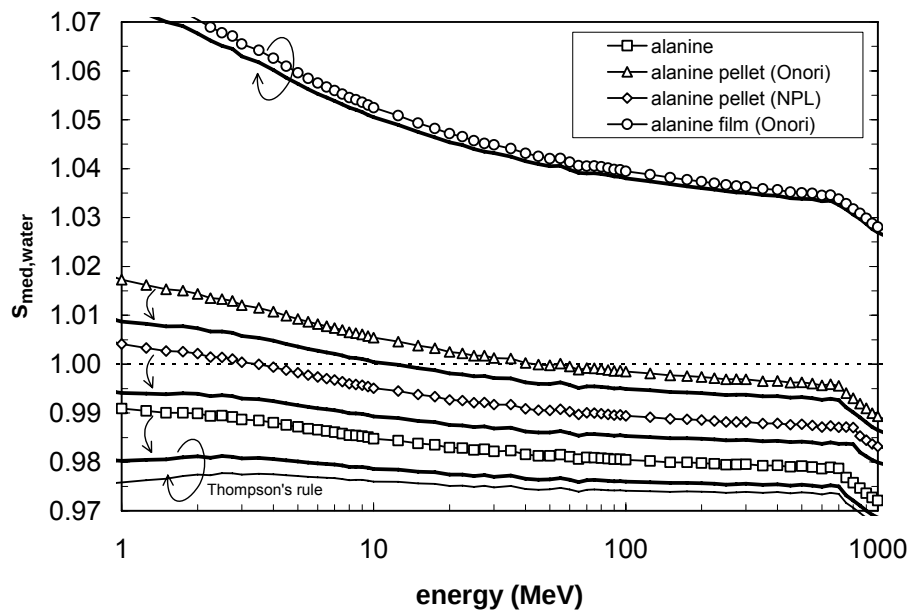


Figure 12. Stopping power ratios medium to water for various alanine materials as a function of proton energy. The symbols represent the calculations according to Bragg's rule, the solid lines those with I -values consistent with ICRU report 37.

Figure 12 presents the medium to water stopping power ratios $S_{\text{med,water}}$ for alanine, alanine pellets used by Onori et al. (1997), alanine pellets used by NPL and alanine film used by Onori et al. (1997). The relevant properties of these four materials are presented in table 9. At this point, all materials are considered to be a homogeneous mixture. The curves with symbols represent the values obtained from Bragg's rule and the correction for the density effect. They are converted in the solid curves (as indicated by the arrows) when a correction according to equation (11) is applied using mean excitation energies calculated according to ICRU report 37. For pure alanine an addition curve is shown which was calculated in the same way but with the I -value obtained from Thompson's rule (see ICRU report 37). The 1% difference for alanine, and an almost comparable difference for the other materials, illustrates that it is important to take consistent I -values into account. Figure 13 presents similar data as figure 14 but with PMMA as reference material.

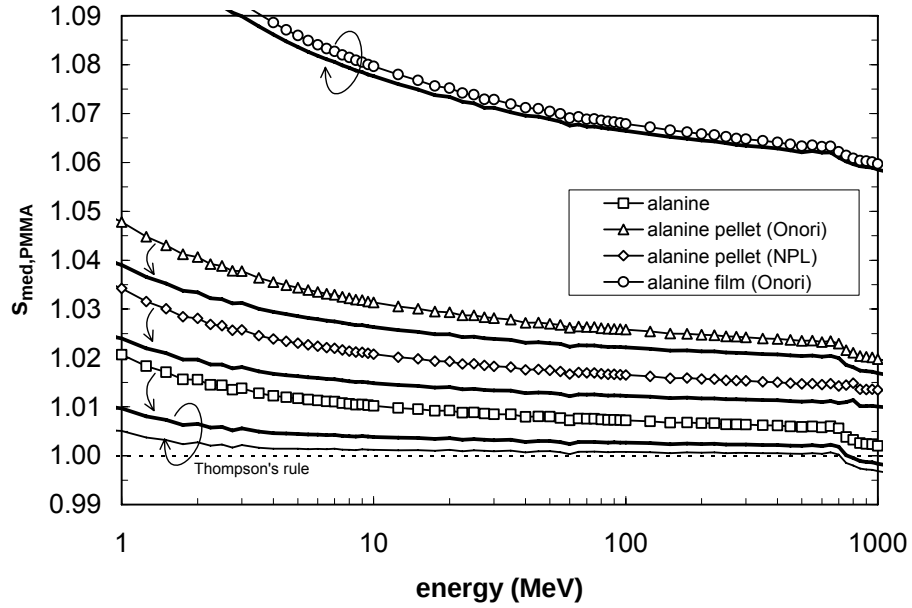


Figure 13. Stopping power ratios medium to PMMA for various alanine materials as a function of proton energy. The symbols represent the calculations according to Bragg's rule, the solid lines those with I-values consistent with ICRU report 37.

The stopping powers calculated above are required as input for condensed history Monte Carlo simulations and will determine the electromagnetic energy loss of a proton during transport. For dosimetry purposes, the previous becomes a bit more complicated when the microscopic structure of the alanine pellets is taken into account. Microscopic pictures of the different stages of the NPL pellets reveal that after grounding, the powder consists of alanine grains with dimensions varying from a couple of μm to $100\ \mu\text{m}$ and paraffin wax grains of similar size. After the heat treatment, clumps of 200 to $500\ \mu\text{m}$ tend to be formed in which the alanine grains are bound together with paraffin. Thus, the dose deposited in the alanine grains will consist of two contributions: one from electrons generated in the alanine grain itself and a second one from electrons generated in the surrounding matrix. For large grains, the former contribution will dominate, whereas for small grains, the latter will. Given the relatively small amount of paraffin, it is reasonable to assume that the surrounding medium is a homogeneous mixture of alanine and paraffin.

This way, Spencer-Attix cavity theory could be applied to relate the dose in the mixture D_{mix} to the dose in the alanine grain D_{al} :

$$D_{\text{mix}} = D_{\text{al}} \cdot s_{\text{mix,al}}^{\text{SA}} \quad (12)$$

or

$$D_{\text{al}} = \frac{D_{\text{mix}}}{s_{\text{mix,al}}^{\text{SA}}} \quad (13)$$

where $s_{\text{mix,al}}^{\text{SA}}$ is the Spencer-Attix stopping power ratio mixture to alanine, which accounts for the delta electrons escaping from the alanine cavity. For non-modulated proton beams, this could be approximated with the restricted mass stopping power ratio with an electron

energy cut-off that is related to the size of the cavity (alanine grain in this case), as shown by Palmans and Verhaegen (2002).

If we assume that the proton fluence is not altered by the presence of the alanine grain, we can assume that D_{mix} is proportional to the fluence by a factor equal to the mass stopping power S_{mix} of the mixture and that D_{al} is proportional to the fluence by a factor which we call here the apparent stopping power $S_{al}^{apparent}$, which is thus given by

$$S_{al}^{apparent} = \frac{S_{mix}}{S_{mix,al}^{SA}} \quad (14)$$

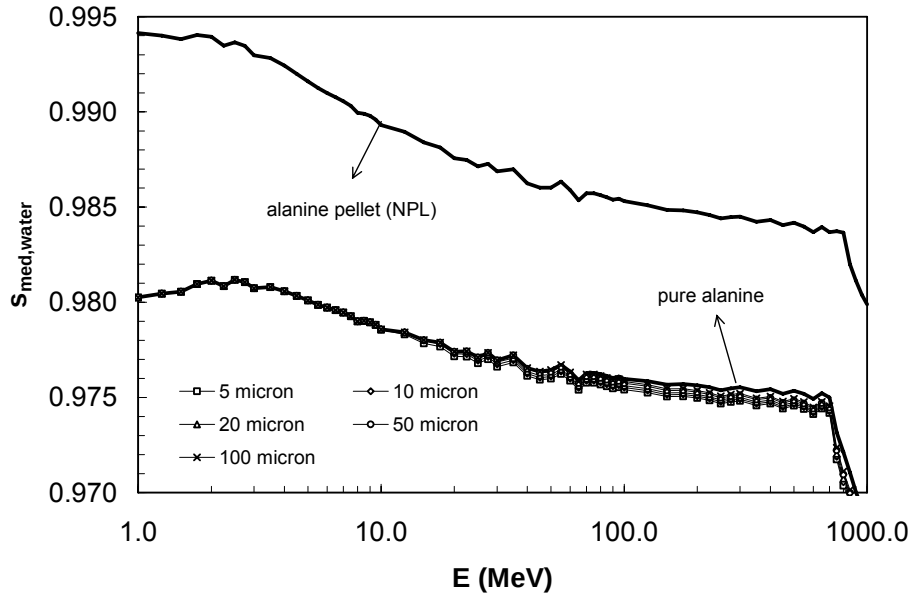


Figure 14. The solid lines give the stopping power ratios medium to water for alanine and NPL's alanine pellets. The symbols give the ratio of the apparent stopping power as defined in the text for different alanine grain sizes.

The ratio of this apparent stopping power to the stopping power of water is plotted in figure 14. The electron cut-off energies are derived from the grain size, such that an electron with that cut-off has a range in the continuous slowing down approximation equal to the average cord length of the grain, which is assumed to be spherical. Surprisingly and contra-intuitively, at lower cut-off energies the values tend to decrease instead of going towards the values of the mixture. It is clear that for cut-off energies corresponding to grain sizes above 5 μm , the energy deposition in alanine will be determined by the alanine stopping power itself and only very slightly (less than 0.1%) by the composition of the mixture.

3.2.1.2. Non-elastic nuclear interactions

Beside electromagnetic interactions with electrons and the nuclei, non-elastic nuclear interactions play an important role in the transport of protons through matter and are a substantial alternative way of energy transfer and deposition. Figure 15 shows the percentage of the proton energy transfer in water, which goes to non-elastic interactions.

About 1/5 to 1/2 of this energy fraction goes to non-charged particles (neutrons and photons) and thus escapes from the geometry, whereas the rest goes to charged particles which deposit their energy relatively close to the location of the energy transfer.

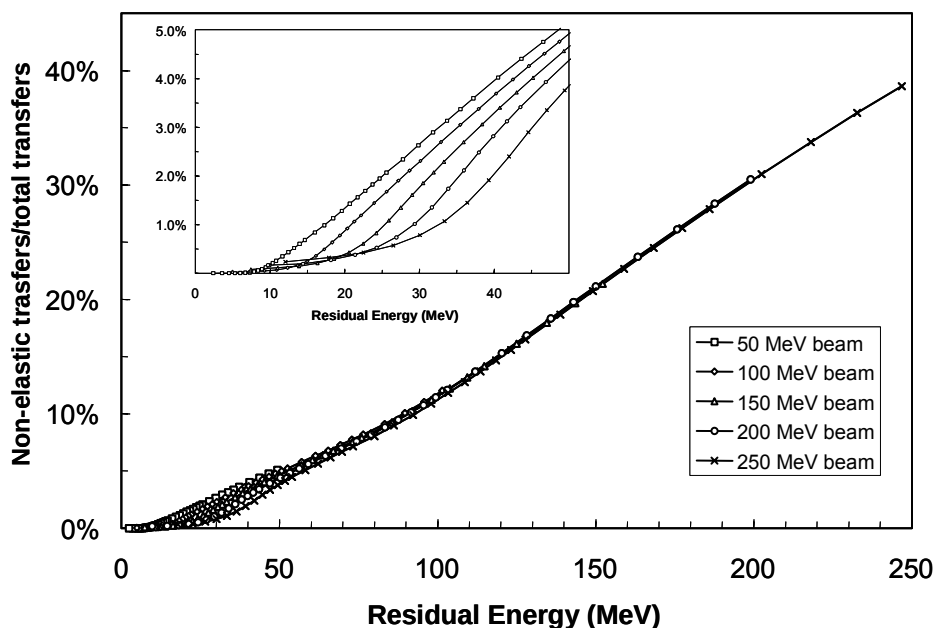


Figure 15. Fraction of the energy transfer in water, which goes to non-elastic nuclear interactions as a function of the residual energy (calculated as defined by Vynckier et al. 1994). The inset magnifies the data from 0 to 50 MeV.

It is clear that especially at higher energies non-elastic nuclear interactions play an important role. Palmans et al. (2002b) showed that the water-equivalence of a medium is entirely determined by the differences in these nuclear interactions. When converting dose in one medium to dose in another medium (e.g. water), the fraction of the dose deposition due to non-elastic nuclear interactions should be converted separately from the electromagnetic contribution. The latter should be converted with the ICRU 49 stopping power ratios, whereas the former should be converted with the ratios of non-elastic nuclear cross sections (for generation of charged secondaries) per nucleon. The more, the different cross sections cause differences in nuclear attenuation from one material to another, such that the fluence at equivalent depths becomes different and fluence correction factors are required (Palmans et al. 2002b).

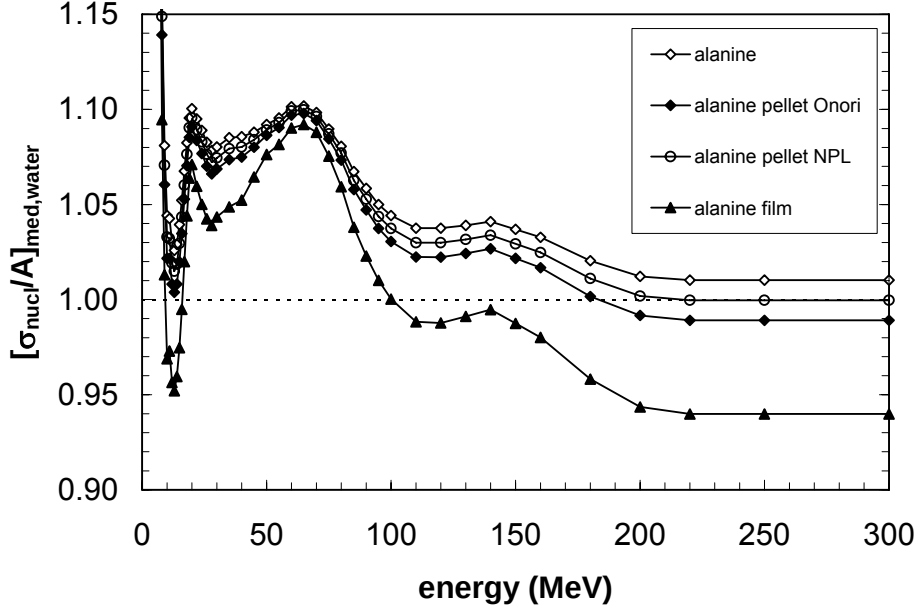


Figure 16. Ratios of total non-elastic nuclear interaction cross sections per nucleon relative to water based on the data from ICRU report 63.

The nuclear interaction cross section per nucleon, (σ/A) , for a material is calculated as

$$\left(\frac{\sigma}{A}\right) = \frac{\sum_i n_i \cdot \sigma_i}{\sum_i n_i \cdot A_i} \quad (15)$$

where n_i is the number of atoms of the i 'th constituent present in an amount of the material (a molecule or a certain mass) and A_i is the atomic weight of that constituent. σ_i are the total nuclear interaction cross sections for the constituent atoms, which are in this work taken from ICRU report 63 (2000).

In figure 16 the ratios of total nuclear interaction cross sections per nucleon are plotted relative to water, for the various alanine-based materials investigated in this study. Figure 17 presents similar data relative to PMMA. These figures show that the non-elastic nuclear interaction cross sections for alanine and the alanine pellets are closer to PMMA than to water. From this point of view PMMA would be a more suitable phantom material than water. This advantage could be slightly counteracted by the closer agreement of the stopping powers with those of water than with those of PMMA. Apart from this, the use of water is not favourable due to the argument that alanine pellets should be sealed from water, which would introduce additional perturbations that have to be accounted for.

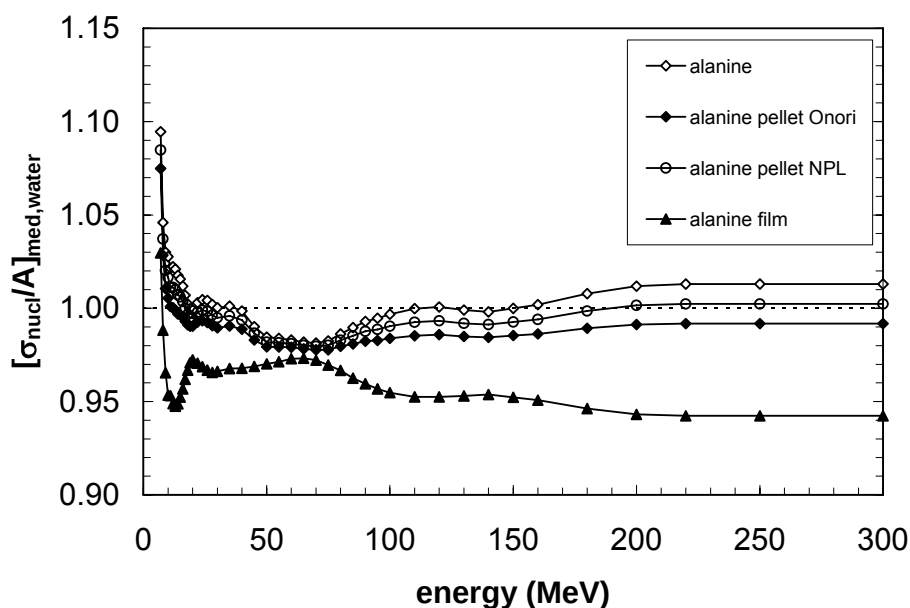


Figure 17. Ratios of total non-elastic nuclear interaction cross sections per nucleon relative to PMMA based on the data from ICRU report 63

3.2.3. Experimental results

The alanine dosimetry service of NPL has been described by Sharpe and Sephton (2000). The alanine pellets consist of a mixture of 90% alanine and 10% paraffin wax. They have a diameter of 5 mm and a nominal thickness of 0.5 mm or 2.5 mm. Readout is performed with a Bruker ESP 300 X-band 9" magnet.

For depth dose measurements in the ocular proton beam, a PMMA phantom was constructed as shown in figure 18. Irradiations were done for modulated and non-modulated beams. In the regions with rapidly varying gradients (the Bragg peak for the non-modulated beam and the distal edge of the modulated beam) the 0.5 mm pellets were used.

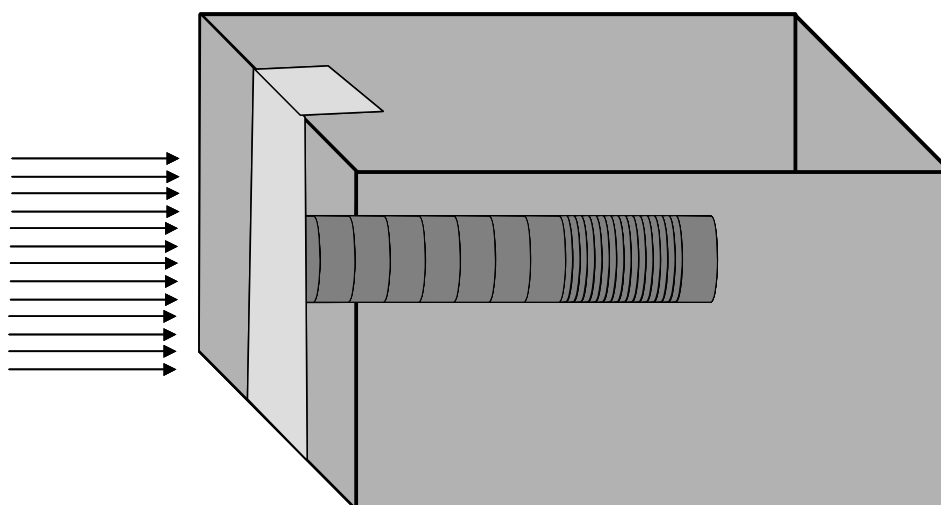


Figure 18. Schematic drawing of the phantom for depth dose measurements using alanine pellets

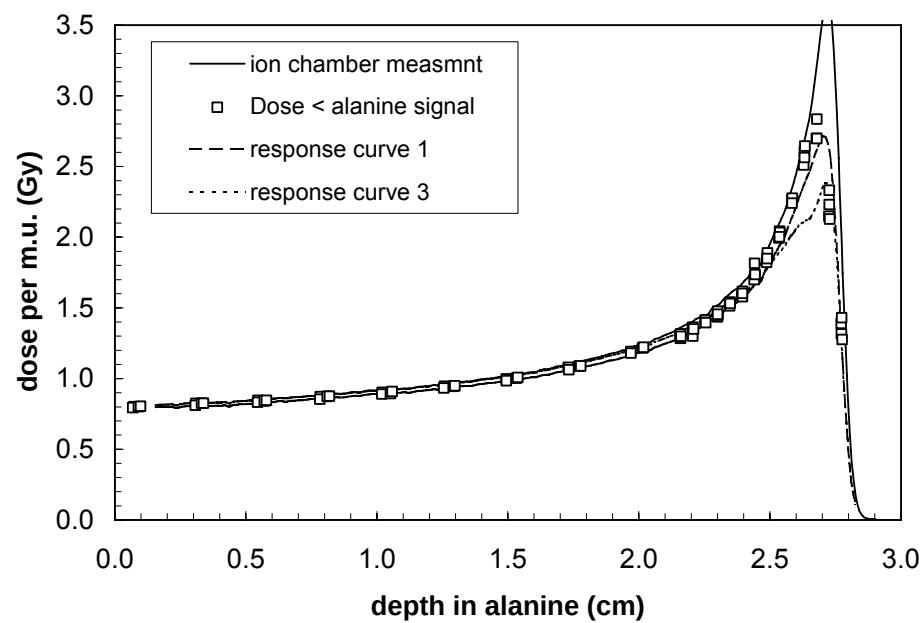


Figure 19. Dose per monitor unit measured with alanine pellets in non-modulated proton beam compared with Markus chamber measurement (applying IAEA TRS-398).

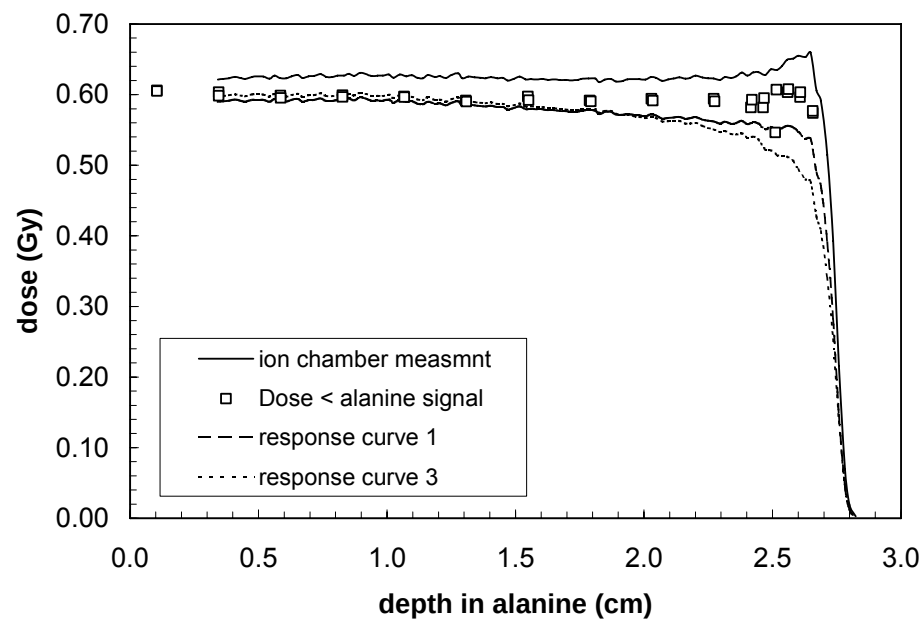


Figure 20. Idem as figure 19 for modulated proton beam.

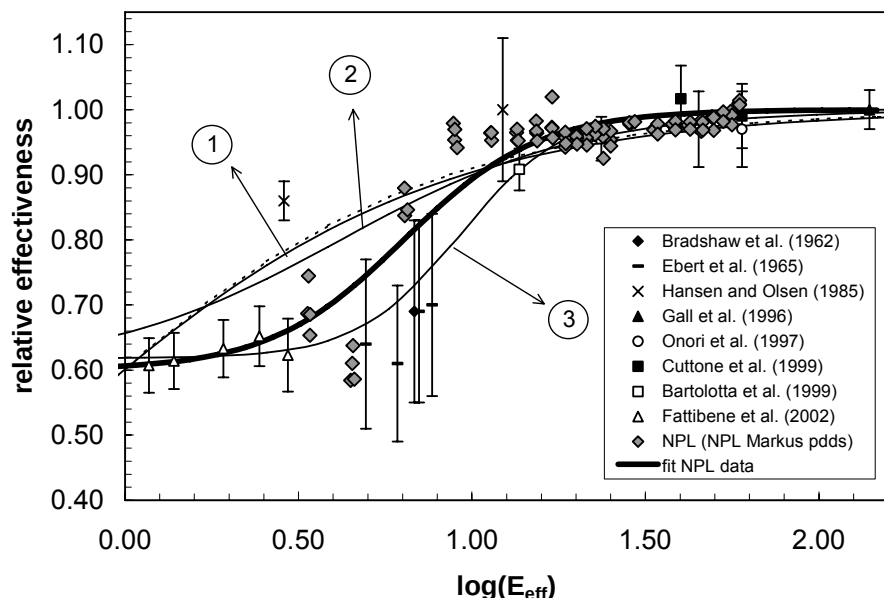


Figure 21. Data from figure 11, together with the data extracted from the measurements with the NPL pellets in the non-modulated beam. A fit to the NPL data with a lower energy saturation of 0.6 is shown as well.

Figures 19 and 20 show the measured depth dose response curves. The results are absolute dose per monitor results with the alanine response based on a ^{60}Co calibration, so no normalisation has been done. In the non-modulated beam, four sets of data are available, in the modulated beam two. It is obvious that the response in the high-energy region (low depths) agrees well with the ionisation chamber measurements but that in the Bragg peak there is a clear under response. For the modulated beam this results in a reduced response over the whole range of depths, consistent with the low-energy component present in the SOBP.

Figure 21 plots the relative effectiveness derived from the measurements in the non-modulated beam (taken as the ratio of alanine dose over ion chamber dose) together with the data from the literature. The data are consistent within the uncertainties. A proper uncertainty analysis of the NPL results has not been done yet but the uncertainties on the lower energy data points are considerable since it is not sure what exactly the effective energy is in the Bragg peak. A sigmoidal fit performed to the NPL data points lies more or less in between the response curves 1 and 3. This seems to contradict the results in the modulated beam but it needs to be investigated how important the low-energy component is as a function of depth. There is a wide region where the NPL data points are above the three predicted response curves.

A possible way to investigate this further would be to irradiate pellets in very low energy protons such as from the Van de Graaff generator in the neutron section of NPL.

3.3. Ionisation chamber measurements

3.3.1. Comparison of two calibration chains for ionisation chambers

This work has been published by Thomas et al. (2003) and is reproduced in addendum C.

3.3.2. Recombination correction measurements

Highly accurate measurements of the recombination correction for ionisation chambers were performed in July 2004 and repeated in November 2004 and January 2005. The set-up that allows to achieve an very high correlation between test chamber and monitor chamber is shown in figure 22

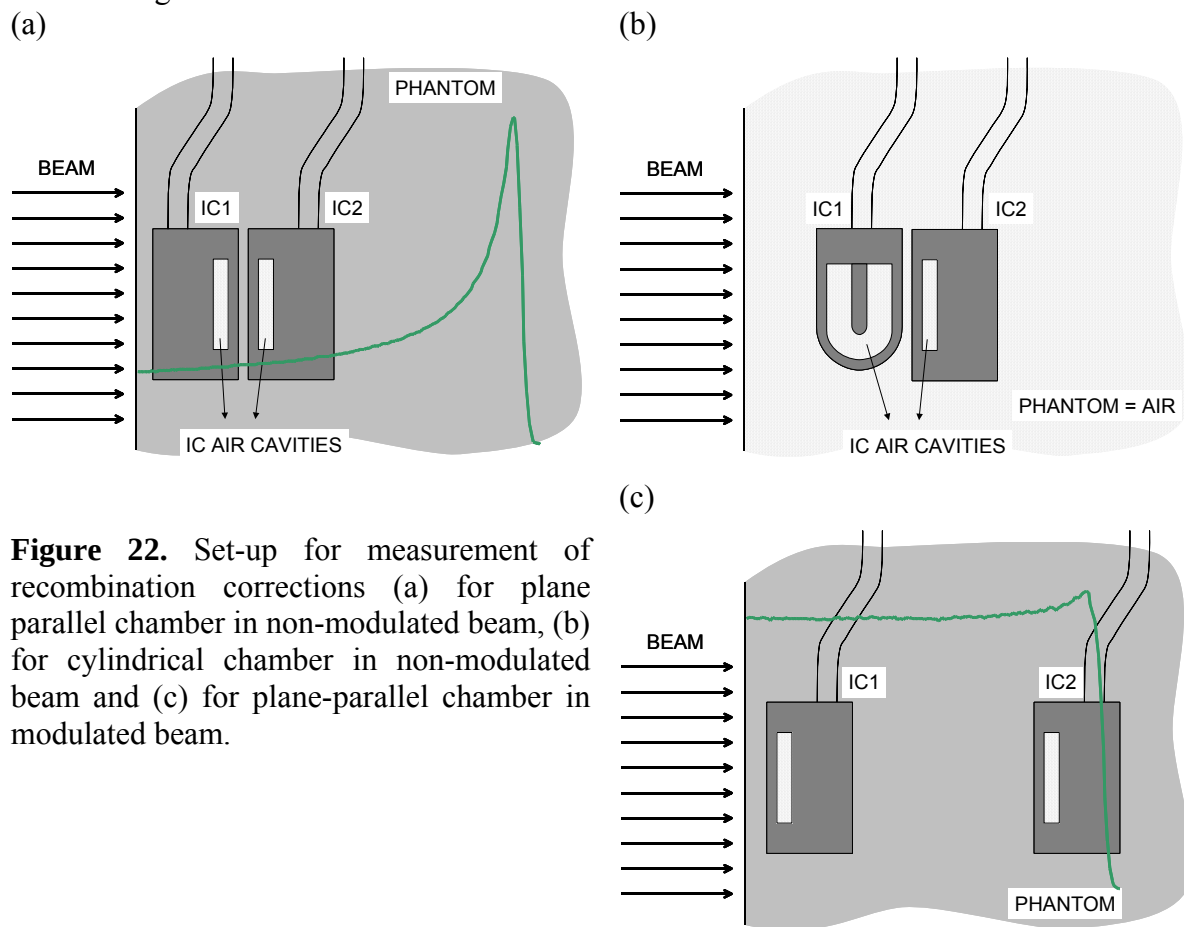


Figure 22. Set-up for measurement of recombination corrections (a) for plane parallel chamber in non-modulated beam, (b) for cylindrical chamber in non-modulated beam and (c) for plane-parallel chamber in modulated beam.

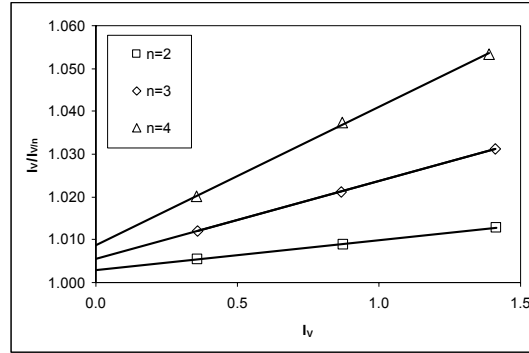
The theoretical derivation of the expressions needed to derive the initial and volume recombination parameters is discussed in addendum F, which reproduces a paper prepared on these measurements (Palmans et al. 2006). The results of the ionisation current at three voltage divisions and three different dose rates is shown for all the individual chambers in figures 23 and 24.

Figure 23 shows the results for the non-modulated beams. The average value of m^2 as derived from the model of equation (20) is $(3.5 \pm 0.4) \cdot 10^{14} \text{ s m}^{-1} \text{ C}^{-1} \text{ V}^2$ that is in good agreement with the value of Boutillon (1998), which was $(3.97 \pm 0.07) \cdot 10^{14} \text{ s m}^{-1} \text{ C}^{-1} \text{ V}^2$.

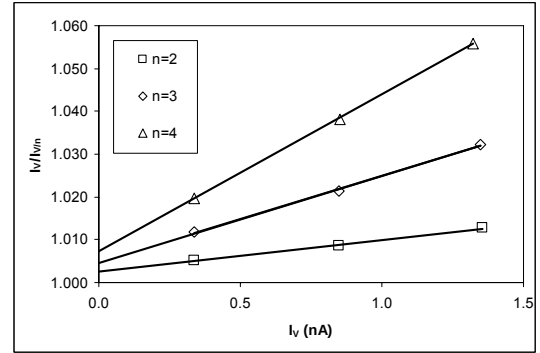
The first set of results for the modulated beams showed some problems. The correlation between the ionisation chamber signals might not have been that good given the separation between the two chambers. There was also a lot of beam instability during the whole of this measurement but especially for the $n=3$ highest dose rate point (boldly encircled in the graph) the dose rate was drifting dramatically. Depending on the assumption that was made

on when this drift started, this point scatters all over the place. For the present analysis it was assumed to be an average over the drift prior and after this measurement.

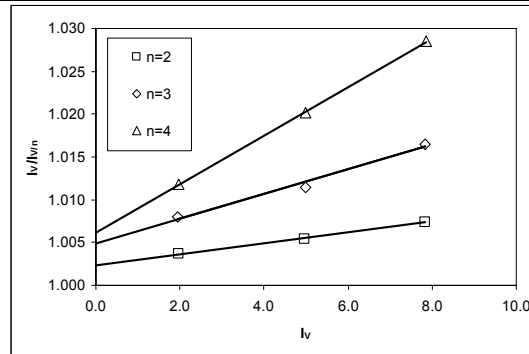
Markus SN478 with Markus SN2225 as monitor



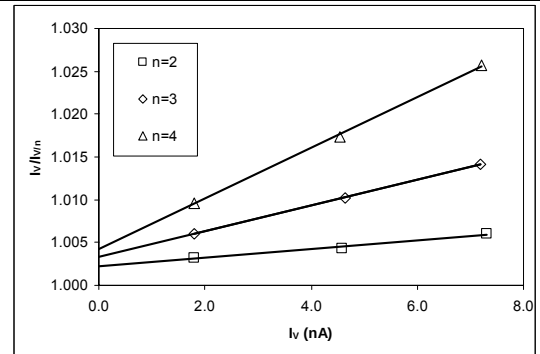
Markus SN2225 with Markus SN478 as monitor



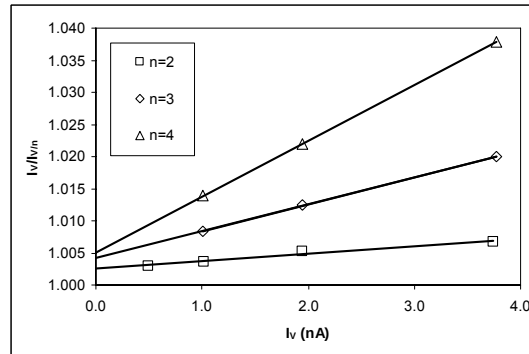
Roos SN133 with Roos SN227 as monitor



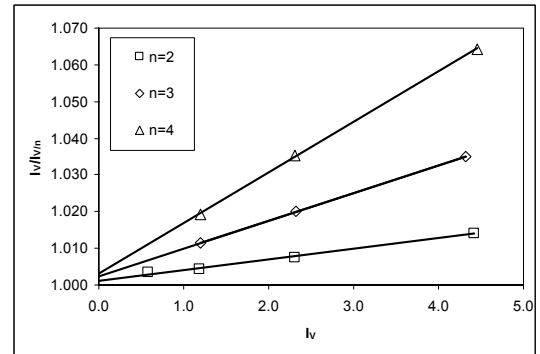
Roos SN227 with Roos SN133 as monitor



NACP02 SN5205 with NACP02 SN8508 as monitor



NACP02 SN8508 with NACP02 SN5205 as monitor



NE2561 SN193 with NACP8508 as monitor

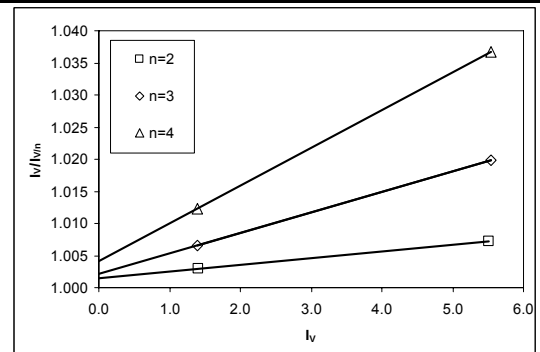
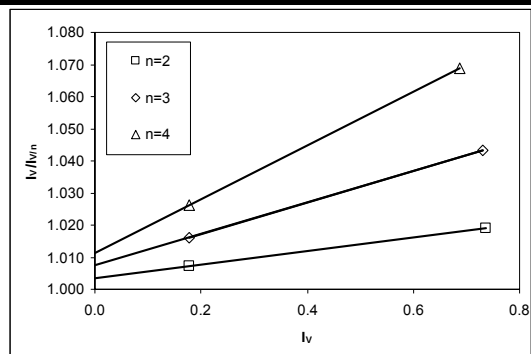


Figure 23. Results of the recombination measurements in the non-modulated beam.

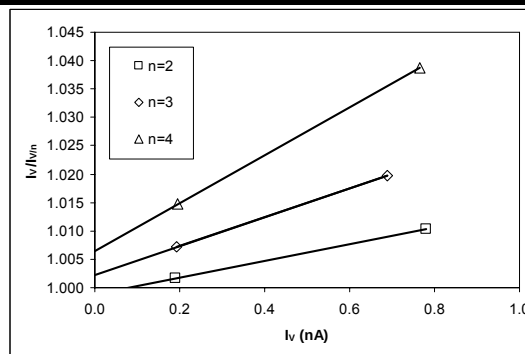
Later sets of measurements in the modulated beam, also shown in figure 24, using the set-up of figure 22a yielded very good results. It should be taken into account in the analysis that the time-averaged current is not the immediate current of importance for the recombination effect. In addendum F it is explained how to take this into account by assuming an

equivalent effective current. There is an indication for a slight LET dependence of the initial recombination

Markus SN478 at 24 with Markus SN2225 at 5 as monitor



Markus SN2225 at 5 with Markus SN478 at 24 as monitor



Markus SN2225 at 5 with Markus SN478 at 24 as monitor - alternative analysis (assuming initial recombination from full-energy measurement)

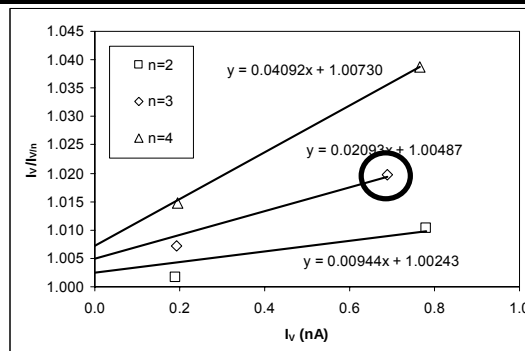
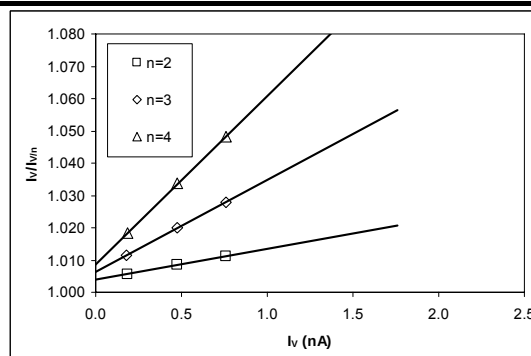


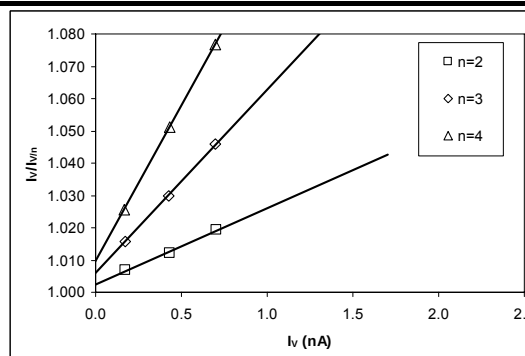
Figure 24. Results of the recombination measurements in the modulated beam.

The four figures below show results at various depths in the modulated beam.

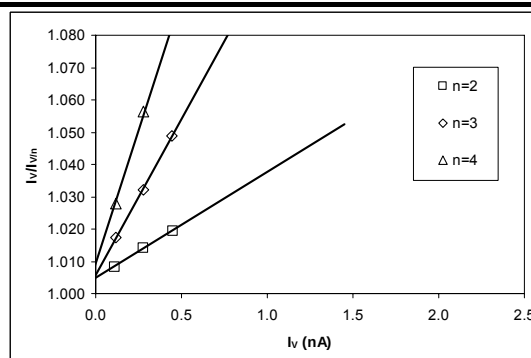
Markus/478@Rres=12.8mm with Markus/2225 as monitor



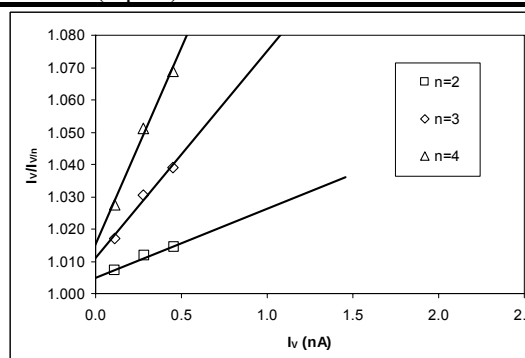
Markus/478@Rres=3.5mm with Markus/2225 as monitor



Markus/478@Rres=0.7mm with Markus/2225 as monitor



Markus/478@Rres=0.7mm with Markus/2225 as monitor (repeat)



3.4. Relative dosimetry (1D, 2D and 3D)

3.4.1. 2D lateral film measurements were compared with diode measurements.

Figure 25 gives an appreciation of how a 2D lateral profile looks like. Figure 26 shows 1D lateral profiles derived at the X- and Y-axis of these film measurements for modulated and non-modulated beams. These profiles have to be compared with those in figure 7, measured with a diode. Again, it is obvious that the dip in the profile is more pronounced for the non-modulated beam than for the modulated beam. The beam profile in the X-direction is also the one, which is more difficult to control, which results in the tilted profiles.

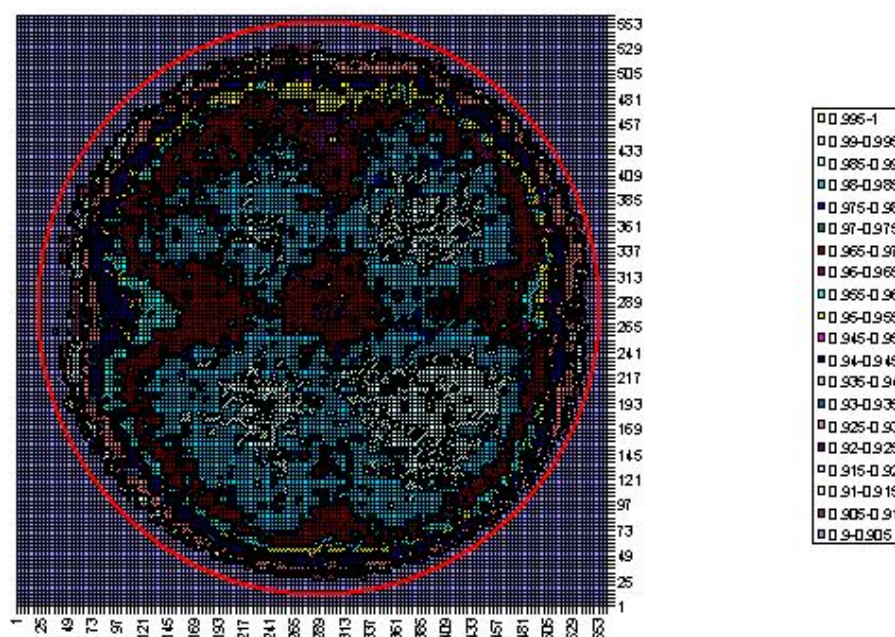


Figure 25. 2D lateral profile in a modulated beam at the phantom surface measured with radiographic film.

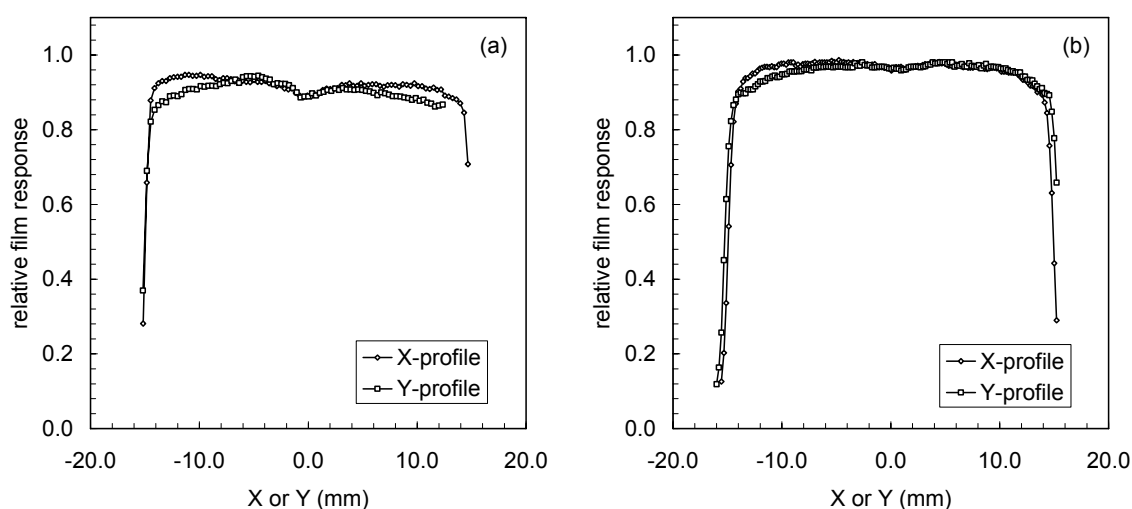


Figure 26. 1D profiles on the X- and Y-axis derived from film measurements (a) for the non-modulated beam and (b) for the modulated beam.

3.4.2. 3D dose distributions with normoxic gel dosimeters

3D dose distributions with normoxic gel dosimeters were obtained in collaboration with Yves De Deene and Koen Vergote from Ghent University, Belgium to be compared with diode and ionisation chamber measurements. Figure 27 gives depth dose curves measured with two different chemical gel compositions. The depth dose curves were obtained in a magnetic resonance imaging scanner at Ghent University using specially developed imaging sequences and a calibration of the gels in a 6 MV high-energy x-ray beam. We can see that there is a distinct difference in the response between the two gel compositions. The underlying chemistry combined with LET-dependence of the proton energy of the chemical yields is at present under investigation.

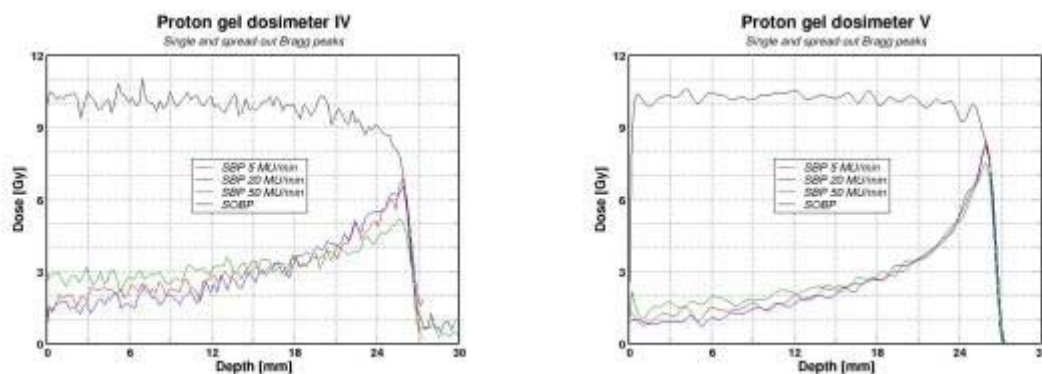


Figure 27. Depth dose distributions measured with two different normoxic gel compositions in a modulated beam and for three different dose levels in the non-modulated beam.

4. SIMULATIONS IN THE SCOPE OF THE LIGHT ION DOSIMETRY PROJECT

A number of Monte Carlo (MC) simulations and heat transport simulations were performed in support of experiments or interpretation of experimental results. An overview of the Monte Carlo codes that were used will be given and the development of the in-house Monte Carlo codes McPTRAN.CAVITY and McPTRAN.RZ based on the existing PTRAN code will be outlined. Comparisons between the various codes and applications to beam line simulations and dosimetry problems will be given.

4.1. Monte Carlo

In the next sections a brief description of the Monte Carlo codes used is given in alphabetical order, followed by the applications for which they were used.

4.1.1. *GEANT4*

“This is a toolkit for the simulation of the passage of particles through matter written in C++. Geant4 provides a complete set of tools for all the domains of detector simulation: Geometry, Tracking, Detector Response, Run, Event and Track management, Visualisation and User Interface. An abundant set of Physics Processes handle the diverse interactions of particles with matter across a wide energy range, as required by Geant4 multi-disciplinary nature; for many physics processes a choice of different models is available.

In addition a large set of utilities, including a powerful set of random number generators, physics units and constants, Particle Data Group compliant Particle management, as well as interfaces to event generators and to object persistency solutions, complete the toolkit.

Geant4 exploits advanced Software Engineering techniques and Object Oriented technology to achieve the transparency of the physics implementation and hence provide the possibility of validating the physics results.”

(Excerpt from the geant 4 website:

<http://wwwasd.web.cern.ch/wwwasd/geant4/geant4.html>)

4.1.2. *MCNPX*

This is a general purpose Monte Carlo radiation transport code written in Fortran 90 (but also fortran 77 compatible) that tracks all element particles up to GeV range energies. The version used here, MCNPX 2.4.0, combines MCNP4C3, a coupled neutron-photon-electron Monte Carlo transport code system for energies up to 20 MeV and MCNPX 2.3.0 the previous version of MCNPX, LAHET 2.8 a code system for high energy particle transport calculations and CEM a cascade excitation model for nuclear reactions by the Monte-Carlo method.

Geometric cell boundaries are usually defined by first- and second-degree surfaces.

Transport is simulated using a class I Monte Carlo scheme. Multiple scattering of protons is Goudsmit-Saunderson, energy straggling is Vavilov. For nuclear interactions it uses the LA150H proton data tables for 41 isotopes, the LA150N neutron data tables for 42 isotopes and the LA150U photonuclear data for 12 isotopes. All these data are tabulated from 1 to 150 MeV except the neutron data for Be-9, which only goes to 100 MeV. For higher energies intermediate interaction physics models are used.

Various variance reduction techniques are used and the user interface is through the input file where numerous tallies can be specified for control of transport parameters, selection of alternative libraries or models and detailed data output for analysis.

For the lower energy region of neutrons and photons point wise continuous-energy cross section data are used, It contains intermediate interaction physics includes models for - Intranuclear Cascade Models - Multistage Pre-equilibrium Models (MPM) - Fermi-Breakup Model - Evaporation Model - Level Densities - High-Energy Fission - High-Energy Interactions contains an early version of the FLUKA high-energy code. As to Nuclear Data Tables the following approach is used: For low energy neutrons, tabular data is used from continuous energy libraries as for MCNP.

(Information taken from <http://mcnpx.lanl.gov/documents.html>)

4.1.3. SRIM

SRIM is the abbreviation of “Stopping and Range of Ions in Matter”. It consists of a group of programs, which calculate the stopping, and range of ions (10 eV - 2 GeV /amu) into matter using a quantum mechanical treatment of ion-atom collisions (we shall always refer to the moving atom as an "ion", and all target atoms as "atoms"). It is mainly used for penetration of ions in solids with emphasis on the radiation damage.

The transport algorithm used is TRIM (Transport of Ions in Matter), which tracks ions in a slab geometry, calculating the energy transferred in every target atom collision. During transport simulations it will calculate 3D dose distributions and kinetic phenomena associated with energy losses: target damage, sputtering, ionization, and phonon production. All target atom cascades in the target are followed in detail. This program also uses a multiple interaction algorithm between major collision events.

During collisions, the ion and atom have a screened Coulomb collision, including exchange and correlation interactions between the overlapping electron shells. The ion also has long-range interactions with target atoms creating electron excitations and plasmons within the target. These are described by including a description of the target's collective electronic structure and interatomic bond structure when the calculation is set-up. The charge-state of the ion within the target is described using the concept of effective charge, which includes a velocity dependent charge state and long range screening due to the collective electron sea of the target. A full description of the calculation is found in the tutorial book "The Stopping and Range of Ions in Solids", by J. F. Ziegler, J. P. Biersack and U. Littmark, Pergamon Press, New York, 1985 (new edition in 2003). This book presents the physics of ion penetration of solids in a simple tutorial manner, then presents the source code for SRIM programs with a full explanation of its physics. Further chapters document the accuracy of SRIM and show various applications.

(Information taken from <http://www.srim.org/>)

4.1.4. PTRAN3D, McPTRAN.MEDIA, McPTRAN.CAVITY, McPTRAN.RZ

The McPTRAN-codes are the only set of in-house developed codes based on the transport algorithm of PTRAN3D and these are discussed in more detail below. Overviews and more detailed discussions are also given by Palmans et al. (2004c, 2005).

4.1.4.1. PTRAN3D

PTRAN3D is a class I Monte Carlo code for proton transport in the energy range of clinical proton beams written in fortran 77. Its application is restricted to the simulation of pencil beams in homogeneous water and the calculation of depth dose data, radial dose and fluence deposition data and spectral distributions as a function of depth. McPTRAN.MEDIA, McPTRAN.CAVITY and McPTRAN.RZ are in-house developed codes that use the transport algorithm of PTRAN but extend the capabilities to simulate transport in other materials than water, through inhomogeneous (including dynamically changing) geometries and for proton beams with a variety of characteristics.

PTRAN (Berger 1993) uses the Monte Carlo method to simulate the transport of proton beams through water. Details on the use of the programs and the methods used in the Monte Carlo transport are given by Berger (1963). A brief summary is given in this section.

The incident proton energies for which the code is designed range from 50 to 250 MeV, although it is applicable to a wider range of energies. PTRAN contains several cross section preparation programs and two main codes, PTRAN3D and PTRAN1D. The code take into account multiple scattering and Coulomb interaction energy loss mechanisms along with non-elastic nuclear interactions. Deposition of energy as a function of depth and radial distance from the beam as well as energy spectra of the primary protons as function of depth are scored.

The simulations follow a class I condensed random walk scheme (Berger 1963). in which each proton is transported down to a cut-off energy by dividing its track in a series of short steps. They are based on a pre-calculated energy grid at which the various cross section and probability distributions are evaluated. The following distributions are used:

- (i) Energy losses in Coulomb collisions with atomic electrons are sampled from the Vavilov energy straggling distribution (Vavilov 1957) using the ICRU report 49 stopping powers (ICRU 1993) as average values.
- (ii) Multiple scattering deflections due to elastic scattering by atoms are sampled from the Molière distribution (Molière 1948).
- (iii) Energy losses in non-elastic nuclear interactions are based on fits to experimental data (Renberg et al. 1972 and Carlson et al. 1975) based on theoretical considerations (Seltzer 1993).

A number of support programs are available for pre-calculating the required distributions:

* *PARAM* calculates grids and various parameters:

- (i) An energy grid with energy steps $\Delta T_i = T_i - T_{i+1}$ which are either a constant value ΔT or $k \cdot T_i$ (whichever is smaller) down to a chosen cut-off energy T_{cut} .
- (ii) The path length for each energy interval in the continuous slowing down approximation (CSDA) using ICRU 49 stopping powers.
- (iii) Parameters for calculating the Vavilov distribution at T_i .
- (iv) Parameters for calculating the Molière distribution at $(T_i + T_{i+1})/2$.
- (v) Nuclear absorption coefficients, representing the probability per unit path length of a non-elastic nuclear interaction, are calculated from the non-elastic nuclear cross sections. From these, CSDA survival weight factors are derived as a function of energy T_i , which thus represent the fraction of protons (on the average) that has not undergone a non-elastic nuclear interaction when slowed down from energy T_0 to T_i .

* *VPREP* calculates the Vavilov distribution for each grid energy T_i . The data are prepared for enabling sampling using the alias sampling method.

* *MPREP* calculates the Molière distribution for each grid energy T_i [at $(T_i + T_{i+1})/2$]. The data are prepared for enabling sampling using the alias sampling method.

PTRAN applies the following transport scheme:

- (i) All protons start at $(x,y,z) = (0,0,0)$ in the direction of the z -axis (polar axis) with an energy E_1 which can be specified at the start of the program.
- (ii) In the n^{th} step, the proton starts with an energy E_n and a direction characterised by a polar angle θ_n and an azimuthal angle φ_n , both with respect to the initial direction of the protons at the 1st step (laboratory coordinate system).
- (iii) The energy loss $E_n - E_{n+1}$ is sampled from the Vavilov distribution at the energy grid value T_i that is nearest to E_n .
- (iv) The path length Δs_n is calculated by interpolating to the energy E_n in the path length versus T_i table.
- (v) The multiple-scattering angular deflection is sampled by sampling an angle θ' from the Molière distribution for the energy grid interval $[T_i, T_{i+1}]$ and φ' from a uniform distribution between -180 and $+180$ degrees. θ' and φ' are both specified in a coordinate system $x'y'z'$ whose polar axis coincides with the direction of the proton at the beginning of the step and are converted to angles θ_n and φ_n at the end of the step by a rotational transformation.
- (vi) The displacements in the coordinate system $x'y'z'$ are calculated and transformed into displacements Δx , Δy and Δz in the laboratory coordinate system by the same rotational transformation.
- (vii) If the energy E_{n+1} is lower than the cut-off energy E_{cut} or a final energy E_{fin} , defined upon starting the calculation, the remaining energy of the particle is binned in a track-end array.

The scoring geometry is very simple: a set of scoring planes is defined perpendicular to the direction of the incident proton beam as shown in figure 28. The depths z_i of these planes are defined in terms of the CSDA range r_0 . In each plane, an array of concentric radial bins with centre $(x,y) = (0,0)$ is defined for the scoring of radial energy deposition distributions as well as an array of energy bins for the scoring of spectra. The positions of the scoring planes and the radial and energy bins are defined in a boundary input file.

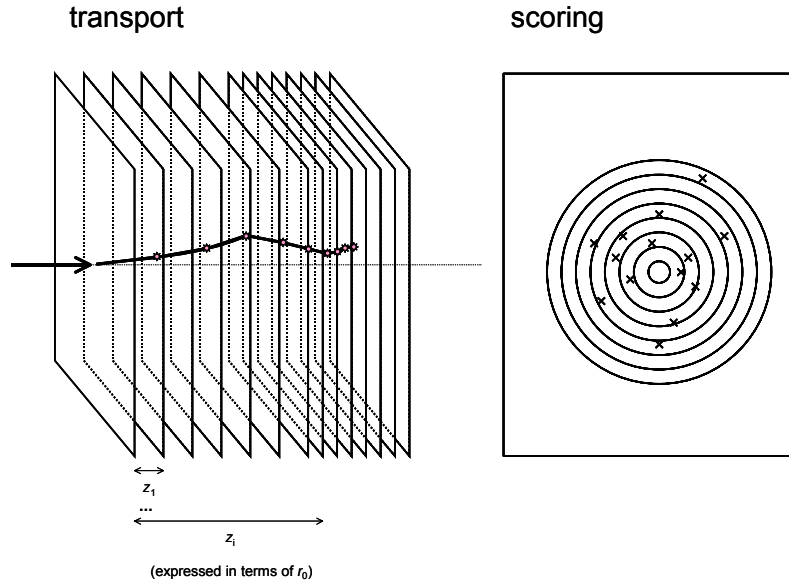


Figure 28. Scoring geometry of PTRAN3D

If a step during the transport simulation of a proton traverses one of the defined scoring planes, the energy E_{cross} , the lateral position $(x_{\text{cross}}, y_{\text{cross}})$ and the polar angle θ_{cross} are determined by linear interpolation between the conditions at the beginning and at the end of the step. The survival factor W_{cross} , the stopping power S_{cross} and the nuclear absorption factor μ_{cross} are then evaluated at the energy E_{cross} . The following quantities are then scored:

- (i) the energy loss in Coulomb interactions, $(dE/dz)_C$, estimated as $S_{\text{cross}} \cdot W_{\text{cross}} / \cos \theta_{\text{cross}}$,
- (ii) the energy loss in non-elastic nuclear interactions, $(dE/dz)_N$, estimated as $E_{\text{cross}} \cdot \mu_{\text{cross}} \cdot W_{\text{cross}} / \cos \theta_{\text{cross}}$,
- (iii) the fluence estimated as $1 / \cos \theta_{\text{cross}}$.

The fluence is binned in the energy spectra and $(dE/dz)_C$ in the radial bins. Since the energy transferred in non-elastic nuclear interactions is only partially going to charged particles (the rest is escaping through neutrons and photons) this fraction can be taken into account in a post-processing program PTSUM. In this, the energy transferred to secondary charged particles is deposited at the position where they are generated. This is a crude approximation, which is not accurate for high-energy protons where secondary proton disequilibrium has a significant effect on dose distributions.

Two random number generators are foreseen with PTRAN:

- (i) A congruential generator which requires only two statements:

```
IR=IAND(MASK,IR*MULT)
R=RNORM*IR
```

where $\text{RNORM} = 2^{-31} = 4.656612873\text{E-}10$ and $\text{MASK} = 2147483647$

PTRAN3D uses two random number sequences with different MULT values the period of the sequence is about $5.37 \cdot 10^8$

- (ii) A lagged Fibonacci random number generator with a very long period (about 10^{144}) but which is considerably slower than the congruential generator

In the simulations performed in this work, only the lagged Fibonacci generator is used.

4.1.4.2. *McPTRAN.MEDIA*

McPTRAN.MEDIA is a modified version of PTRAN3D, which allows the simulation of

- (i) other materials than water
- (ii) inhomogeneous slab geometries
- (iii) broad and non mono-energetic beams
- (iv) the dynamic feature of a modulator wheel.

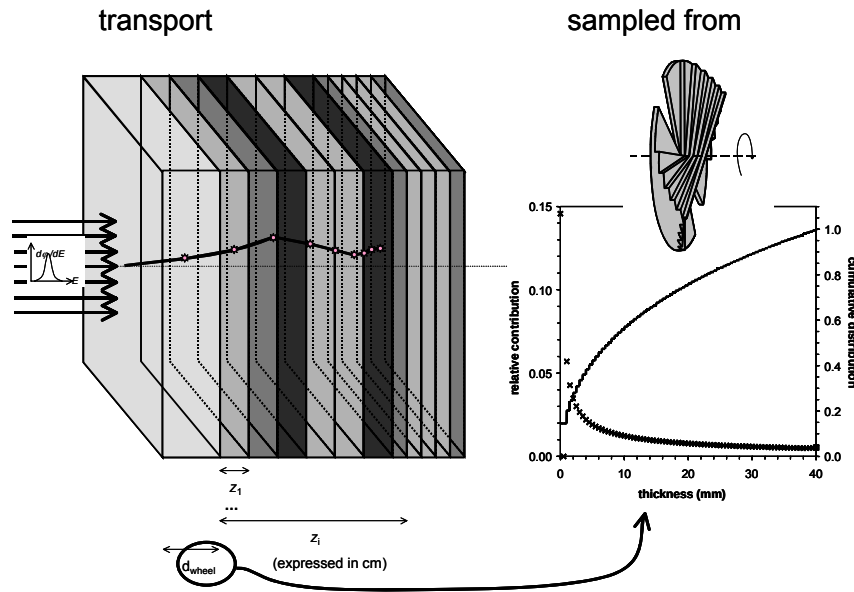


Figure 29. Scoring geometry of McPTRAN.MEDIA

The implementation of other materials than water is mainly a matter of including the data for those materials:

- (i) For most materials of interest, stopping powers can be taken directly from ICRU report 49. For materials not listed in there, Bragg's additivity rule can be applied although a recent paper (Palmans 2003) shows that some refinements are required in order to be consistent concerning the calculation of the mean excitation energy I_0 and the Barkas correction.
- (ii) The calculation of the Vavilov distribution require the first moments S_1 and I_1 of the oscillator strength distribution and the mean excitation energies for the calculation of a small correction to the theory introduced by Shulek et al. (1966). S_1 and I_1 were taken from Inokuti et al. (1978, 1981) for all elements with Z ranging from 1 to 38. Bragg's rule was applied for compounds.
- (iii) For the calculation of the Molière multiple scattering distributions, a correction factor k_{HF} has to be applied to the screening angle. These data were taken from Berger and Wang (1988).
- (iv) Non-elastic nuclear interaction cross sections were taken from Janni (1982) and ICRU report 63 (ICRU 2000) in the studies described below. For the elements not specified in these references, log-log interpolation of the cross sections as a function of the atomic number was used. In recent work, only the ICRU report

63 data have been considered since they form a more comprehensive set. Apart from gathering the required data, some technical modifications to the coding of the cross section preparation programs.

- (v) In addition, the range of incident beam types was extended from only mono-directional and mono-energetic beams to circular beams, rectangular beams, beams with an angular distribution and beams with a Gaussian energy distribution.

For the situation where the slabs in figure 29 can contain different media, a boundary crossing algorithm is required. The technical changes required for this are:

- (i) When a boundary is encountered in a particle's trajectory, the particle is transported to the boundary along the trajectory. The energy loss along the track is scaled with the ratio of the step length up to the boundary to the initially sampled step length. At the boundary, the energy E_{bound} , the polar angle θ_{bound} and the survival factor W_{bound} , are evaluated in the same way as the scoring plane crossing quantities (section 2.3). Given the small scattering angles and energy losses that are involved, we can assume that the errors due to this approximation are small. An evaluation of the influence of the step-cutting artefact revealed no significant effect (Palmans and Verhaegen 1998). For the next step, new multiple scattering angles and energy loss are calculated based on the material properties of the new region.
- (ii) The expression of all distances in cm instead of scaled with r_0 .
- (iii) Giving all material dependent data an additional array dimension.
- (iv) Administration of the media information for each layer defined by two scoring planes.

The implementation of the dynamic geometry of a modulator wheel is, as illustrated in figure 29, done by including an additional layer at the front of the geometry. For each incident proton, the thickness of this layer is sampled from the distribution of the wheel thickness. The incident proton is then set in motion at the front of this additional layer. The usual situation where the wheel is a long distance away from the phantom can be easily dealt with by defining the second layer as air with the thickness equalling the distance between the modulator wheel and phantom.

4.1.4.3. *McPTRAN.CAVITY*

McPTRAN.CAVITY is a modified version of PTRAN3D, which allows the simulation of doses in cavities

The existing scoring algorithm in PTRAN3D is inconvenient for calculating doses in inhomogeneous geometries that are more complex than slab geometries. In McPTRAN.CAVITY the original scoring algorithm was abandoned and cylindrical, spherical and thimble cavity geometries were introduced, with the specific purpose of simulating ionisation chamber perturbations. Essentially, all elements from the existing geometry in McPTRAN.MEDIA were retained. Although the scoring planes have no function in the scoring algorithm any more, they proved to be useful for geometry interrogation reduction. In between two of the existing scoring planes a number of non-intersecting and embedded spherical and/or cylindrical cavities can be defined. The situation of two concentric cylindrical cavities is illustrated in figure 30.

The new scoring algorithm has the following features:

- (i) In each geometrical region, the track length can be scored and binned into an energy spectrum. The boundary-crossing algorithm assumes that track length is distributed evenly along the step.
- (ii) The energy loss is calculated as the difference between the begin and end energy of the step (whether it be a 'normal' or a cut-off step).
- (iii) The non-elastic nuclear energy transfer is scored at the average value of the energies at begin and end of the step.
- (iv) If desired, both quantities in the previous two steps can be binned as a function of energy as well.

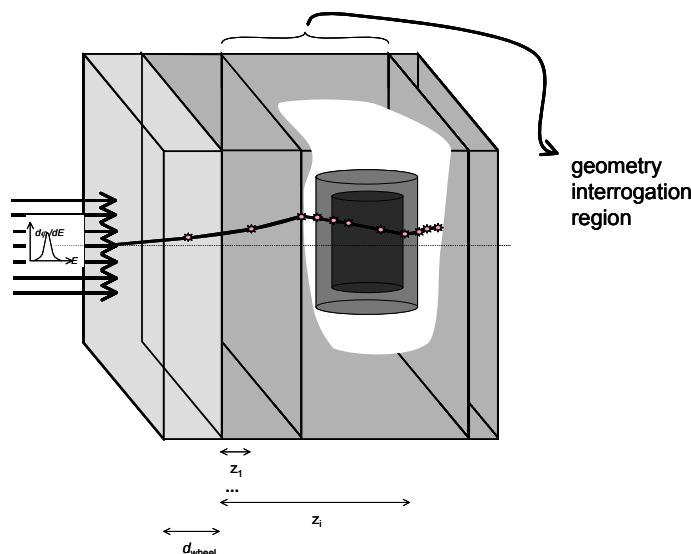


Figure 30. Geometry with dynamic modulator wheel and two concentric cylindrical cavities

A geometry interrogation reduction method is used to avoid unnecessary testing. The test whether or not a particle enters a cavity is only activated in a region between two slabs where the cavities have been specified. This results in a substantial reduction of calculation time.

A correlated-sampling technique was implemented as well, to allow a more efficient calculation of perturbation factors for ionisation chambers with equal geometries but different wall materials. Practically this means that when a particle enters the cavity that defines the outer dimensions of the ionisation chamber wall, the energy, the spatial coordinates and directional coordinates of the proton are stored as well as the random numbers of the simulation at that point. After the transport of that particle is finished, the transport simulation is restarted with the stored conditions but with the wall set to another material. This approach not only saves computing time but also improves the statistical correlation of the doses in the cavity with and without wall (in which case the wall material is set to the surrounding medium).

This code was used to calculate gradient perturbation corrections for all cylindrical ionisation chambers listed in IAEA TRS-398. These calculations were performed on the United Devices distributed computing grid and the results are presented by Palmans (2006a) and Palmans (2006b), reproduced in addendum G.

4.1.4.4. *McPTRAN.RZ*

McPTRAN.RZ is a modified version of PTRAN3D, which allows the calculation of doses in rz-geometry

After all the adaptations for McPTRAN.MEDIA and McPTRAN.CAVITY, the implementation of an rz-voxel geometry is straightforward. The slab geometry is retained and a number of concentric cylinders are constructed perpendicularly to the slabs. In each of the rz-voxels, a different medium can be specified and the same quantities as in McPTRAN.CAVITY can be scored. The geometry is shown in figure 31.

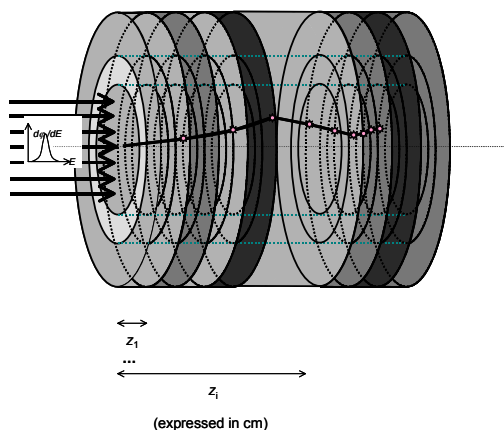


Figure 31. Geometry used in McPTRAN.RZ

4.1.5. *Applications*

4.1.5.1. *Applications of PTRAN*

Figure 32 shows depth distributions of $(dE/dz)_C$ and $(dE/dz)_N$, spectral fluence distributions and radial energy deposition distributions at a few depths for incident proton energies of 60 MeV and 150 MeV beam.

Comparison with similar data obtained with MCNPX, GEANT4 and SRIM are published by Shipley et al (2005) and reproduced in addendum E.

4.1.5.2. *Applications of McPTRAN.MEDIA*

McPTRAN.MEDIA was used to calculate spectral distributions in modulated beams. These could be used to calculate stopping power ratios. It was shown earlier by Palmans and Verhaegen (1998) that water to air stopping power ratios in a modulated beam down to distances of about 1.5 cm from the distal edge of the pdd were not more than 0.3% higher than in non-modulated mono-energetic proton beam at the same residual range R_{res} (the distance from the depth of the distal 10% dose level). This result was used in IAEA TRS-398 (Andreo et al. 2000) to justify the use of a unique beam quality specifier and data set, common to both modulated and non-modulated proton beams. Example spectra in a modulated beam at various depths are shown in figure 33.

McPTRAN.MEDIA was also used to calculate fluence correction factors. TRS-398 (Andreo et al. 2000) recommends to perform reference dosimetry with ionisation chambers in a water phantom. However, when performing dosimetry in low-energy beams, it could be necessary to do the measurements in a plastic or graphite phantom in order to achieve

sufficiently accurate positioning. A fluence correction factor should then be introduced to account for any difference in fluence at equivalent depths in the plastic phantom and in water. This situation can be compared with the one for electron beams. A preparatory study showed that differences in scaled depth dose curves between different materials were almost entirely due to differences in non-elastic nuclear interaction cross sections (Palmans and Verhaegen 1997). McPTRAN.MEDIA was used to evaluate fluence correction factors for PMMA and polystyrene and compare these values with experimental results (Palmans et al. 2002). Later, fluence correction factors were calculated for graphite to examine the water equivalence of graphite in calorimetry by Palmans et al. (2003) in addendum A and for some ICRU tissue specifications to evaluate the importance of incorporating material dependent non-elastic nuclear interaction data in treatment planning by Palmans and Verhaegen (2005) in addendum D.

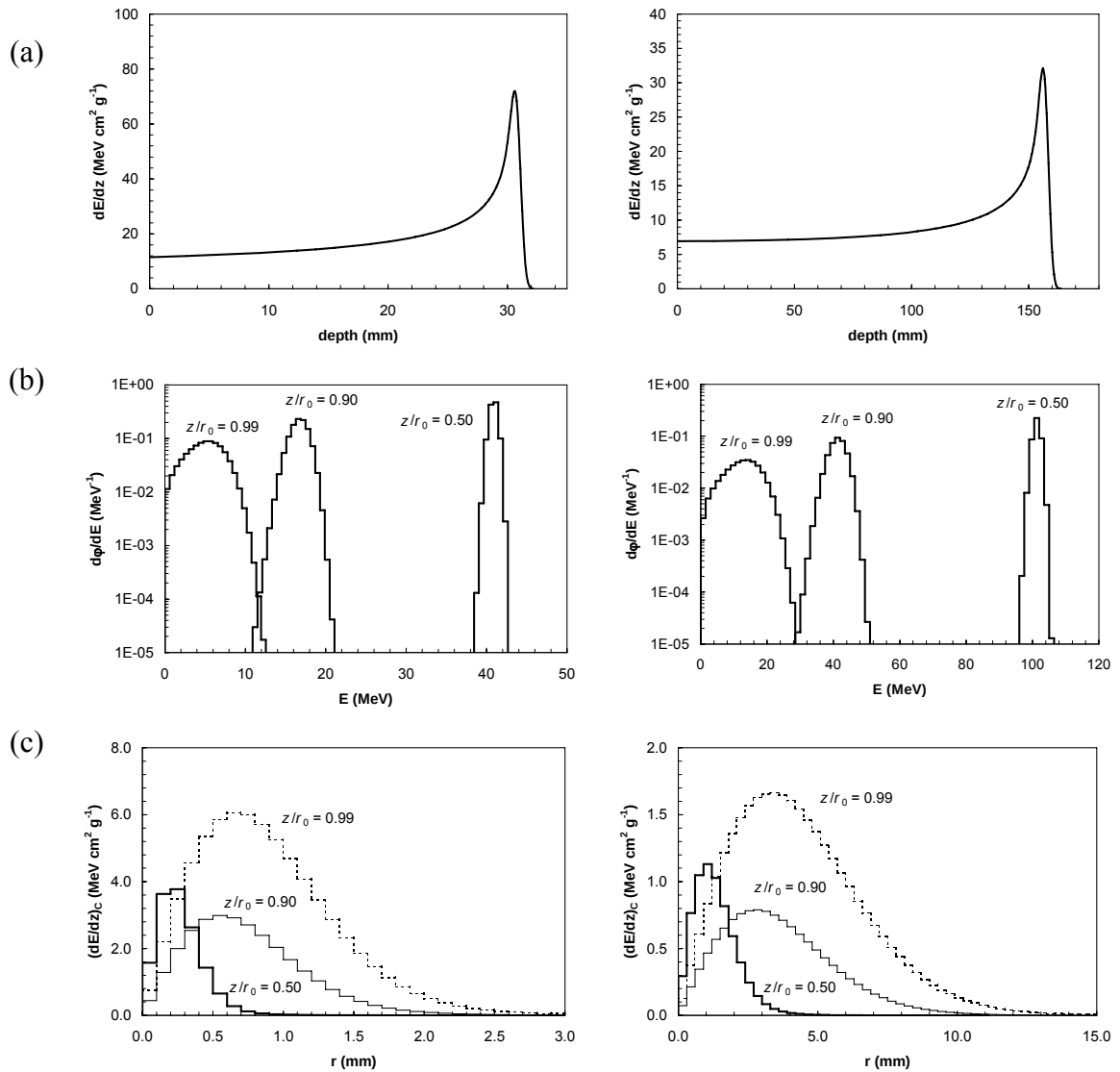


Figure 32. (a) calculated depth dose data with PTRAN3D, (b) spectral fluence distributions and (c) radial energy deposition data for a 60 MeV beam (left) and a 150 MeV beam (right)

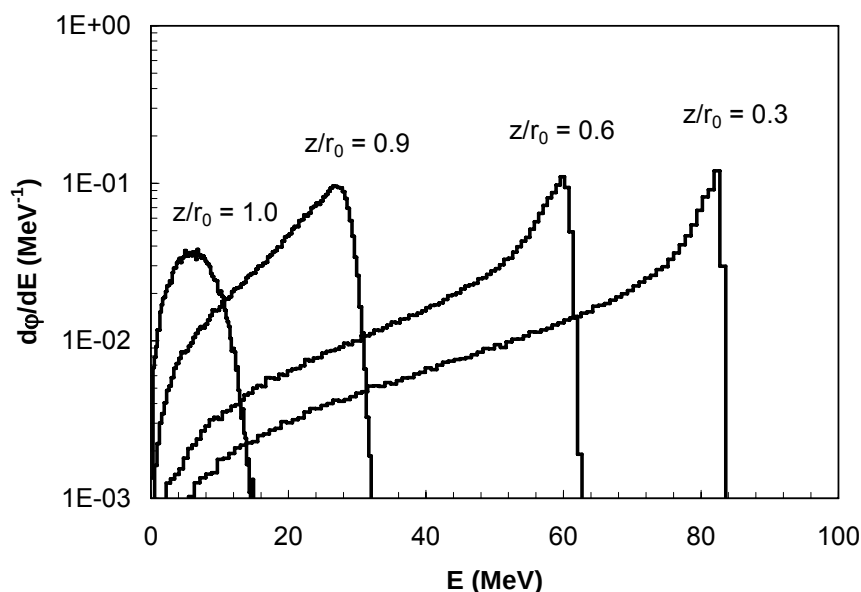


Figure 33. Spectral fluence distributions in a 100 MeV modulated proton beam at various depths in water normalised per incident proton

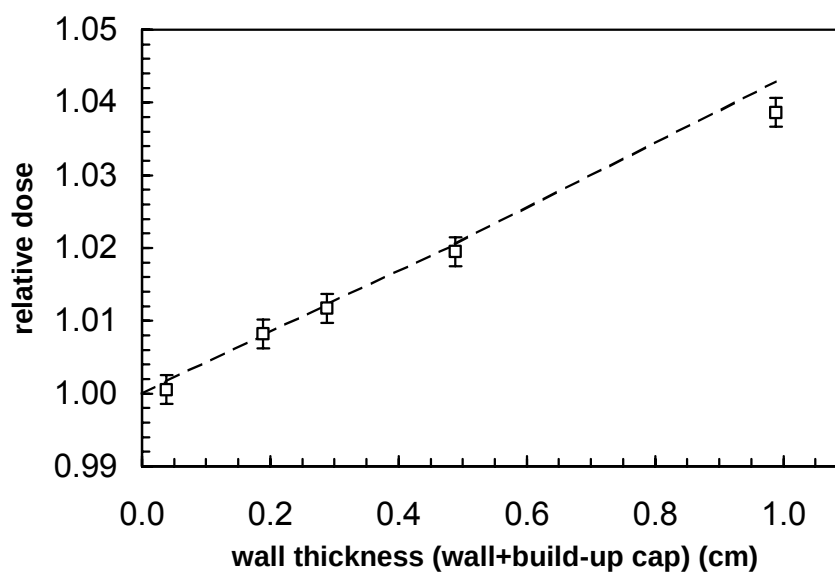


Figure 34. Monte Carlo calculated (dashed line) and measured (symbols) variation of the ionisation chamber signal as a function of graphite wall thickness (reproduced from Palmans et al. 2001)

4.1.5.3. Applications of *McPTRAN.CAVITY*

The response of a graphite walled NE2571 ionization chamber was measured as a function of wall thickness in a PMMA phantom. To this end, four graphite build-up caps (serving as sleeves to increase the apparent wall thickness) and corresponding PMMA holders to fit in the phantom were constructed. The thicknesses of the build-up caps were 1.5 mm, 2.5 mm,

4.5mm and 9.5 mm. The measured dose response as a function of wall thickness was compared with Monte Carlo calculated values using McPTRAN.CAVITY. The results, reported by Palmans et al (2001), are shown in figure 34. At that stage only cylindrical cavities were modelled in the code.

The code was used in a similar to calculate the effective point of measurement for all thimble shaped ionisation chambers listed in IAEA TRS-398 as well as for developing a method for the derivation of a depth dose curve in homogeneous water from the measured depth dose response by the ionisation chamber. These results are presented in addendum G.

The track lengths scored in a volume can be used to evaluate the probability distribution of the generation of δ -electrons with energies above a certain threshold (1 keV in this case). This method was used to calculate the amount of energy that was transported out of the air cavity of an ionisation chamber by electrons and the amount of energy that was imported from the surrounding regions (wall, sleeve and medium) (Verhaegen and Palmans 2001). The Bahba cross-section for the generation of secondary electrons was used (Bahba 1938) and the transport of these electrons was simulated with EGSnrc (Kawrakow 2000). That way, δ -electron perturbation correction factors were calculated for ionisation chambers and the results were compared with experimental data (Palmans et al 2001) shown in figure 35.

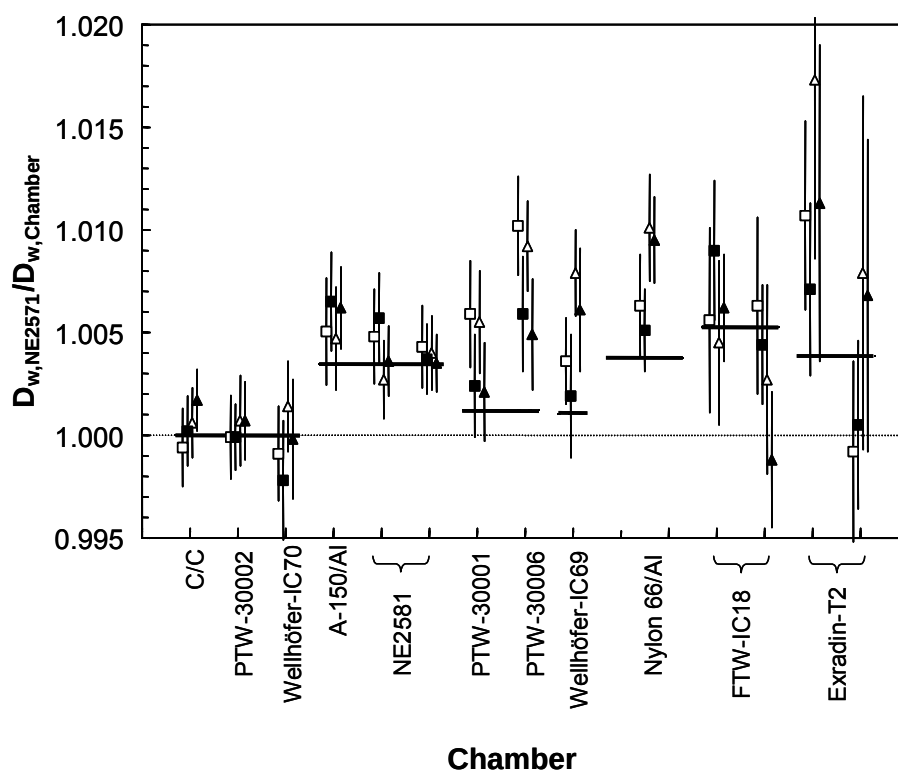


Figure 35. Comparison of the $p_{\text{wall,e}}$ -ratios (relative to the NE2571 chamber) calculated using Monte Carlo (thick horizontal lines) and obtained experimentally in a 75 MeV proton beam. Square symbols are for a non-modulated beam, triangles for a modulated beam, open symbols are based on absorbed dose calibrations and full symbols based on air kerma calibrations (Adapted from Palmans and Verhaegen 2002).

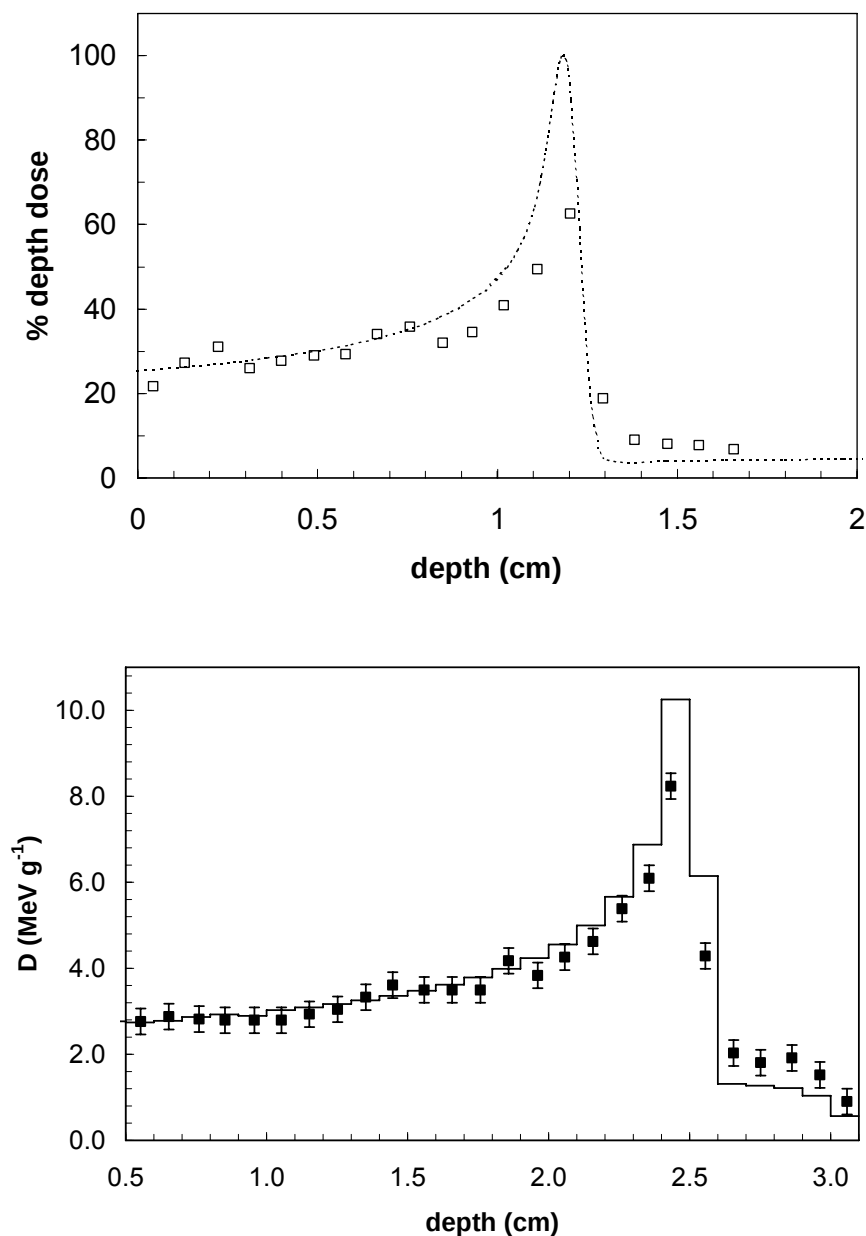


Figure 36. Dose in a stack of TLD pellets (upper) and a stack of alanine pellets in a PMMA phantom, measured (symbols) by Fattibene et al. (1996) for the TLD results and Onori et al. (1997) for the alanine results and calculated with McPTRAN.RZ using a Gaussian energy distribution with a mean value of 62 MeV and a variance of 0.5% (line)

4.1.5.4. Applications of MCNPX, GEANT4 and McPTRAN.RZ to simulation of the CCO beam line

These results have been published by Shipley et al (2005), reproduced in addendum E and by Baker et al (2006).

4.2. Heat transfer simulations

Heat transfer simulations using FEMLAB (Comsol 2003) were performed to determine heat loss perturbations in the SPGC measurements. The results are presented in the paper of Palmans et al (2004a) reproduced in addendum A.

5. ACHIEVEMENTS OF THE SR PROJECT “IMPROVED LIGHT-ION DOSIMETRY” AND CONCLUSIONS

This section copies the information from the final report sent to the science strategy group in October 2004. The previous chapters and addenda contain information that is more recent.

5.1. KEY ACHIEVEMENTS

- A literature survey has been performed on light ion beam dosimetry and two internal reports produced
- Dose distributions have been calculated by Monte Carlo simulations using the codes PTRAN and MCNPX. Work with the codes GEANT4 and SRIM has been initiated.
- Preliminary measurements of relative dose distributions and relative response of ionization chambers have been performed.
- Graphite calorimetry measurements have been compared with ionisation chamber measurements. A value of $(w_{\text{air}})_p$, the average energy required to produce an ion pair in air, is derived from these measurements, which is an important contribution to the very limited amount of data available and will influence future evaluations of the $(w_{\text{air}})_p$ value. The work will thus have a substantial impact on ionisation chamber dosimetry for clinical proton beams in the future.
- Dosimetry using cylindrical and parallel-plate ionisation chambers based on two calibration routes has been compared. The results will have an impact on future dosimetry recommendations.
- The energy dependence of the alanine dosimeter has been investigated based on the literature and by experiments. The work demonstrated that alanine could be used for relative dosimetry and as dosimeter for a potential quality assurance service.

5.2. PROGRESS TOWARDS OBJECTIVES

A literature survey of about 200 papers on the dosimetry of light-ions has been performed.

Measurements at Clatterbridge Centre for Oncology (CCO) have been performed. Depth dose distributions in water and graphite were measured with plane-parallel ionisation chambers and compared with diode measurements. The water-equivalence of graphite has been evaluated by measuring fluence correction factors for these ionisation chambers. Stacks of alanine pellets in a PMMA phantom were irradiated.

A literature survey on the basic interaction data and the LET dependence of Alanine has been performed. An assessment of the stopping power of Alanine materials as a function of energy and composition has been made. Non-elastic nuclear interaction cross-sections assessed for Alanine.

Three proton Monte Carlo codes have been installed at NPL (GEANT4, PTRAN and SRIM) and one at McGill University (MCNPX).

Stopping power and nuclear data for alanine have been implemented in PTRAN as well as LET-dependent response functions. Depth dose curves in homogeneous media have been calculated (water, graphite, PMMA, alanine). Depth dose curves in homogeneous water calculated with PTRAN, MCNPX and SRIM have been compared.

Monte Carlo simulations and analytical calculations have been performed to assess gradient correction factors for cylindrical ionisation chambers.

Absolute dose measurements with cylindrical and plane-parallel ionisation chambers in 'plastic water', in water and in graphite have been compared, based on two calibration routes: calibrations in Co-60 and in a 19 MeV high-energy electron beam. The results show that the four ionisation chamber types (NE2561, NACP02, Markus, Roos) perform and agree well in an ocular proton beam. There is however a systematic difference in the dosimetry between both calibration routes, which is of concern since for plane-parallel chambers a calibration in electron beams, is generally recommended whereas for cylindrical chambers a calibration in a ^{60}Co photon beam is generally recommended. With the present information we cannot rule out whether the problem is due to the calibrations or due to the data in the dosimetry protocols and argues for more experimental work in proton dosimetry.

Graphite calorimetry measurements were performed with an existing portable calorimeter but were problematic due to large heat flows. A dedicated small body portable graphite calorimeter (SPGC) was constructed and tested in the proton beam with satisfactory results. The SPGC response was then compared with ionisation chambers in two 3-day runs which allowed us to derive a value for $(w_{\text{air}})_p$, the average energy required to produce an ion pair in air, which is an essential quantity in ionisation dosimetry. The results have been presented at a specialised Calorimetry Workshop, at the World conference on Medical Physics and Biomedical Engineering and are accepted for publication in a peer-reviewed journal.

Absolute measurements with alanine pellets in water have been compared with ionization chambers with good and reproducible agreement when taking the alanine energy response into account. From measurements of depth dose curves with alanine pellets in a plastic phantom we derived information on the energy response of alanine. It has also been demonstrated that it is possible to load a phantom with alanine pellets, send it to the proton centre for irradiation and back by mail which makes it suitable as a quality assurance service tool. An experiment to investigate the effect of the air gaps between the phantom and the pellets is at present being analysed.

Compared dosimetry practice of Clatterbridge Centre of Oncology (CCO) with our dosimetry (traceable to portable graphite calorimeter)

Detailed measurement of recombination correction factor for NACP02 plane-parallel chamber

Initiated collaboration with Ghent University on the use of normoxic gels in proton dosimetry. 10 gels with different chemical compositions were irradiated in mono-energetic beams (three different dose rates) and a modulated beam.

Last set of alanine measurements analysed

A comparison of depth dose curves, energy spectra and radial distributions in water, graphite and alanine using PTRAN, SRIM, GEANT4 and MCNPX has been performed. For low-energy (ocular) proton beams the differences are minor but for high-energy beams (which are used for treating deep seated tumours) the differences are substantial. The major reason is that the various codes use different inelastic nuclear interaction cross-sections. A study of these differences in tissue compositions has been initiated as well in order to

quantify the effects of inelastic nuclear interactions in Monte Carlo treatment planning simulations.

Fluence correction factors for graphite have been calculated using PTRAN, GEANT4 and MCNPX. These factors are required to convert dose measured in a graphite phantom or calorimeter to dose to water, the clinically relevant medium.

Gap and volume averaging correction factors, required for correcting the dose derived from the graphite calorimetry measurements, have been calculated using PTRAN.

The influence of the alanine response as a function of energy on measured depth dose distributions has been investigated with PTRAN as well as the perturbation due to the alanine holder (phantom or sleeve) and air gaps between holder and alanine pellets.

Heat transfer calculations were performed to determine corrections on the post-irradiation drift curves in the graphite calorimeter

Effective point of measurement has been calculated for 58 commercial ionization chamber types. Wall perturbation correction factors have been calculated for 58 commercial ionization chamber types.

PTRAN codes have been installed on NPL's United Devices grid computing system allowing the calculation of ionisation chamber perturbation correction factors with a 100 times higher speed than on a single PC.

PTRAN and MCNPX modelling of the CCO beam line. GEANT4 modelling of CCO beam line initiated.

PTRAN simulations of spatial distribution of LET and formation of primary chemical species for supporting understanding of response of normoxic gel dosimeter in proton beam.

5.3. STRATEGIC PARTNERSHIPS

The Clatterbridge Centre of Oncology (CCO) was a crucial partner for the experimental work and will continue to be a partner after the project. The collaboration gives us access to their facilities for on the average four weeks per year. This would correspond to a cost of about 60K a year. In the framework of the Monte Carlo simulations we collaborated with McGill University, Montreal, Canada through Dr. Frank Verhaegen and with the University of Liverpool through Dr. Colin Baker, whose contributions by performing MCNPX simulations were highly valuable. The amount of work they have put in would correspond to at least 10K and we will also continue these collaborations.

5.4. FORWARD LOOK AND EXPLOITATION PLAN

We aim for publishing the work on alanine dosimetry as well as the work on the calculation of ionisation chamber perturbation factors in peer reviewed journals since, given the expanding nature of the field, there is an increasingly high international interest in such data (among which from an ICRU committee dealing with technical aspects of proton therapy), which are lacking at present.

The work in this SR project has led to several proposals in the formulation of the 2004-2007 Ionising radiation metrology programme, which is at present accessible on the DTI website for public comment. The proposals comprise the development of a portable proton beam dosimetry standard, measurement of proton stopping power interaction data, measurement

supporting the choice of inelastic nuclear interaction data for dosimetry and direct measurements of dose distributions.

We will remain in further contact with proposals for a high-energy proton therapy facility in the UK (the CASIM project has unfortunately not been approved for funding).

Although the CASIM proposal has come to nothing, the SR project has enabled us to embark on an international collaboration with the University of Louvain-La-Neuve Belgium. They have funding for a research project at the (non-clinical) ion beam facility in GANIL, Caen, France for investigating issues related to ionisation chamber dosimetry and have invited us for collaboration as a partner in this project with calorimetry. NPL's contribution is made as part of the NMS programme. This is an example of co-funding that we hope will open opportunities for future new business by exhibiting our expertise in the field to a wider community. The same group from Louvain-La-Neuve is trying to establish similar contacts with the clinical ion beam facility in GSI, Darmstadt, Germany as well.

There is perhaps a possibility of some (low-volume) commercial exploitation of the calorimeter developed for proton beam measurements. We have a contact at IBA who, aware that we were developing calorimeters for proton and ion beams, has recently asked if we can produce them commercially. There are some technical issues to be explored (related to lack of knowledge of interaction data) that we still have to investigate about using graphite in proton and ions before this possibility could be seriously considered but this enquiry will be followed up.

5.5. PEER REVIEWED PUBLICATIONS RESULTING FROM THE WORK IN THIS SR PROJECT

Palmans H. (2003) Effect of alanine energy response and phantom material on depth dose measurements in ocular proton beams, *Technol Cancer Res Treat.* 2:579-86

Palmans H., Thomas R., Simon M., Duane S., Kacperek A., DuSautoy A. and Verhaegen F. (2004) A small-body portable graphite calorimeter for dosimetry in low-energy clinical proton beams, *Phys Med. Biol.* 49:3737-3749

Palmans H. and Verhaegen F. (2005) Assigning non-elastic nuclear interaction cross sections to Hounsfield Units for Monte Carlo treatment planning of proton beams, *Phys. Med. Biol.* 50:991-1000

In preparation:

Palmans H. (2005) Perturbation factors for cylindrical ionisation chambers in proton beams: I. Corrections for gradients, *Phys. Med. Biol.* (prepared for submission)

Palmans H., Thomas R. and Kacperek A. (2005) Recombination correction in the CCO low-energy proton beam, *Phys. Med. Biol.* (prepared for submission)

5.6. PRESENTATIONS RESULTING FROM THE WORK IN THIS SR PROJECT

Conference papers

Palmans H., Thomas R., Simon M., Duane S., Kacperek A., Seco J., Nutbrown R., Shipley D., DuSautoy A. and Verhaegen F. (2004) Feasibility of graphite calorimetry in a modulated low-energy clinical proton beam *Proceedings of the Workshop on Recent Advances in Absorbed Dose Standards*, ARPANSA, Melbourne, Australia, 19-21 August 2003, 13 pages

Thomas R. Palmans H., Kacperek A. and Duane S. (2004) Low energy proton beam dosimetry with plane-parallel chambers using NPL electron and ^{60}Co calibrations *Proceedings of the Workshop on Recent Advances in Absorbed Dose Standards*, ARPANSA, Melbourne, Australia, 19-21 August 2003 (in preparation)

Lecture notes

Palmans H. (2004) PTRAN, a Monte Carlo code for transport simulation of 50 to 250 MeV protons & McPTRAN.MEDIA, McPTRAN.CAVITY and McPTRAN.RZ, a series of derived codes *NPL workshop on Monte Carlo codes*, National Physical Laboratory, 17-18 March 2004, 14 pages

Conference presentations

Palmans H., Thomas R., Kacperek A., Sharpe P., DuSautoy A. and Verhaegen F. (2003) Applying the NPL alanine dosimetry service to low-energy clinical proton beams *Abstracts of the 38th meeting of the Particle Therapy Co-Operative Group*, Chester, UK, 14-16 May 2003

Thomas R., Palmans H. and Kacperek A. (2003) Low energy proton beam dosimetry with plane parallel chambers using NPL electron and ^{60}Co calibrations *38th meeting of the Particle Therapy Co-Operative Group*, Chester, UK, 14-16 May 2003

Palmans H. and Duane S. (2003) Modelling heat transport in calorimeters for radiotherapy dosimetry *11th conference on the Mathematics of Finite Elements and Applications*, Uxbridge, UK, 21-24 June 2003

Palmans H., Thomas R., Simon M., Duane S., Kacperek A., Seco J., Nutbrown R., Shipley D., DuSautoy A. and Verhaegen F. (2003) Feasibility of graphite calorimetry in a modulated low-energy clinical proton beam *Workshop on Recent Advances in Absorbed Dose Standards*, ARPANSA, Melbourne, Australia, 19-21 August 2003

Thomas R. Palmans H., Kacperek A. and Duane S. (2003) Low energy proton beam dosimetry with plane-parallel chambers using NPL electron and ^{60}Co calibrations *Workshop on Recent Advances in Absorbed Dose Standards*, ARPANSA, Melbourne, Australia, 19-21 August 2003

Duane S., Thomas R., McEwen M., Palmans H. and Simon M. (2003) Development of the NPL portable graphite calorimeter and its operation in hospital radiotherapy departments *Workshop on Recent Advances in Absorbed Dose Standards*, ARPANSA, Melbourne, Australia, 19-21 August 2003

Palmans H., Thomas R., Kacperek A., Sharpe P., Nutbrown R., DuSautoy A. and Verhaegen F. (2003) Alanine dosimetry for low-energy clinical proton beams: basic interaction data and perturbation correction factors *World Congress on Medical Physics and Biomedical Engineering*, Sydney August 24-29, 2003

Palmans H., Thomas R., Simon M., Duane S., Kacperek A., Nutbrown R., DuSautoy A. and Verhaegen F. (2003) Application of the NPL graphite and water calorimeters to low-energy clinical protons *World Congress on Medical Physics and Biomedical Engineering*, Sydney August 24-29, 2003

Palmans H. and Verhaegen F. (2003) Perturbation correction factors in clinical proton beams for cylindrical ionisation chambers tabulated in IAEA TRS-398 due to gradient

effects and secondary electron effects *World Congress on Medical Physics and Biomedical Engineering*, Sydney August 24-29, 2003

Verhaegen F., Palmans H. and Seco J. (2003) A Comparison of Hadron Monte Carlo Transport Codes (PTRAN, MCNPX, GEANT) for Clinical Protons *World Congress on Medical Physics and Biomedical Engineering*, Sydney August 24-29, 2003

Duane S., Thomas R., McEwen M., Palmans H. and Simon M. (2003) Development of the NPL portable graphite calorimeter and its operation in hospital radiotherapy departments *World Congress on Medical Physics and Biomedical Engineering*, Sydney August 24-29, 2003

Palmans H., Shipley D., Nutbrown R. and Verhaegen F. (2004) Inelastic nuclear interactions in Monte Carlo simulations for clinical proton beams *10th UK Monte Carlo user group meeting (MCNEG)*, National Physical Laboratory, 15-16 March 2004

Palmans H. and Verhaegen F. (2004) Importance of non-elastic nuclear interactions in Monte Carlo treatment planning for proton beams *Workshop on Current Topics in Monte Carlo Treatment Planning*, Montreal, Canada, 3-5 May 2004

Palmans H., Thomas R., Simon M., Duane S., Kacperek A., DuSautoy A. and Verhaegen F. (2004) Proton W_{air} value derived from comparing graphite calorimetry with ionisation chamber dosimetry in an ocular beam, *40th meeting of the Proton Therapy Co-Operative Group (PTCOG)*, Paris, France, 15-18 June 2004

Baker C., Kacperek A. and Palmans H. (2004) Monte Carlo simulation of a clinical 60 MeV proton beam line, *40th meeting of the Proton Therapy Co-Operative Group (PTCOG)*, Paris, France, 15-18 June 2004

Shipley D., Palmans H. and Kacperek A. (2004) GEANT4 simulation of an ocular proton beam and benchmark against other Monte Carlo codes, *Submitted for the Monte Carlo 2005 Topical meeting*, Chattanooga, USA, 17-21 April 2005

Palmans H. (2004) McPTRAN.CAVITY and McPTRAN.RZ, Monte Carlo codes for the simulation of proton beams and calculation of proton detector perturbation factors, *Submitted for the Monte Carlo 2005 Topical meeting*, Chattanooga, USA, 17-21 April 2005

5.7. OTHER KT PRESENTATIONS RESULTING FROM THE WORK IN THIS SR PROJECT

H. Palmans chaired a proton dosimetry session at the *38th meeting of the Particle Therapy Co-Operative Group (PTCOG)*, Chester, UK, 14-16 May 2003

H. Palmans gave a seminar on proton dosimetry at NPL on 23 July 2003

H. Palmans gave a presentation on the SR project to Sir Peter Willams on 29 October 2003

H. Palmans gave a presentation on proton dosimetry at the Radiotherapy User's group meeting on 9 December 2003

H. Palmans gave a presentation on the SR project to the RS/RAE group on Monday 12 January 2004

H. Palmans chaired a session on proton Monte Carlo simulations at the *Workshop on Current Topics in Monte Carlo Treatment Planning*, Montreal, Canada, 3-5 May 2004

H. Palmans gave seminars on NPL's role in absorbed base dosimetry, containing a section on proton dosimetry, at Peter MacCallum Cancer Centre, Melbourne, Australia, 22

November 2004, at Royal North Shore Hospital, Sydney, Australia, 25 November 2004 and at Sir Charles Gairdner Hospital, Perth, Australia, 2 December 2004.

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ADDENDUM A: PALMANS ET AL (2004A)

PALMANS, H., THOMAS, R., SIMON, M., DUANE, S., KACPEREK, A., DUSAUTOY, A. and VERHAEGEN, F. A small-body portable graphite calorimeter for dosimetry in low-energy clinical proton beams. *Phys. Med. Biol.*, 2004a, **49**, 3737-3749.

A.1. Abstract

Calorimetry has been recommended and performed in proton beams for some time, but never has graphite calorimetry been used as a reference dosimeter in clinical proton beams. Furthermore, few calorimetry measurements have been reported in ocular proton beams.

In this work we describe the construction and performance of a small-body portable graphite calorimeter for clinical low-energy proton beams. Perturbation correction factors for the gap effect, volume averaging effect, heat transfer phenomena and impurity effect are calculated and applied in a comparison with ionisation chamber dosimetry following IAEA TRS-398.

The ratio of absorbed dose to water obtained from the calorimeter measurements and from the ionisation measurements varied between 0.983 and 1.019, depending on the beam type and the ionisation chamber calibration modality. Standard uncertainties on these values varied between 1.9% and 2.5% including a substantial contribution from the k_Q values in IAEA TRS-398. The $(W_{\text{air}}/e)_p$ values inferred from these measurements varied between 33.6 J/C and 34.9 J/C with similar standard uncertainties.

A number of improvements for the small body portable graphite calorimeter and the experimental set-up are suggested for potential reduction of the uncertainties.

A.2. Introduction

Calorimetry is the preferred method for dosimetry of clinical proton beams as recommended in various reports (Lyman et al. 1986, Vynckier et al. 1991, 1994, ICRU report 59, 1998). A recent code of practice, IAEA TRS-398 (Andreo et al. 2000), recommends dosimetry using an ionisation chamber (IC) with an absorbed dose to water calibration factor in the clinical proton beam or, equivalently, to use a measured beam quality correction factor k_Q . Despite this, no calorimetry based long-term primary standards for protons exist at present.

The feasibility of calorimetry for dosimetry of clinical high-energy proton beams as well as for relatively low-energy beams has been demonstrated. Various calorimeter experiments were performed to evaluate the chemical heat defect in water, graphite and A-150 tissue equivalent plastic (see Palmans and Vynckier, 2002, for an overview). Comparisons of IC dosimetry with water calorimeters (Schulz et al. 1992, Siebers et al. 1995, Palmans et al. 1996) and an A-150 calorimeter (Delacroix et al. 1997) have yielded valuable information on the average energy required to produce an ion pair in dry air by protons. For low-energy proton beams used for the treatment of ocular melanoma, on the other hand, the feasibility of calorimetry has not been explored mainly because of the short range of these beams (less than 3.5 cm) and their limited lateral extension (a few cm in diameter).

In this work, we want to explore the possibility of using a graphite calorimeter for low-energy proton beams. The construction of a small portable graphite calorimeter (SPGC), an adaptation of NPL's existing portable graphite calorimeter (PGC) to the specific situation of a small low-energy proton beam, will be described. The correction factors needed for comparing the SPGC with ICs are investigated and a comparison will be discussed.

A.3. Materials and Methods

A.3.1. Proton beam

The 62 MeV proton beam at Clatterbridge Centre of Oncology (CCO) is generated by a cyclotron and used for the treatment of ocular melanoma. A detailed description of the beam is given by Bonnett et al. (1993).

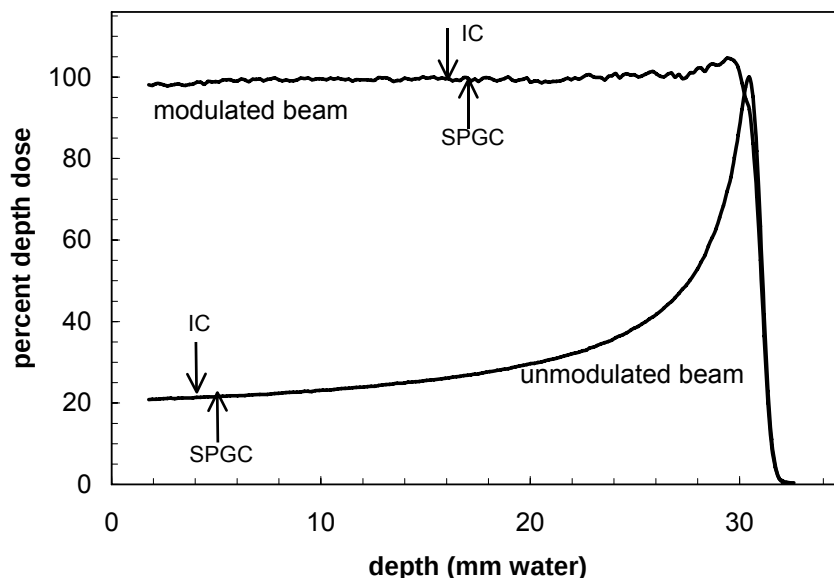


Figure A.1. Depth dose characteristics of the 62 MeV proton beams. The arrows indicate the approximate water equivalent depths at which measurements were performed.

In this work, a 30 mm diameter circular brass collimator was used as a final collimator. A measurement platform was mounted on the positioning mechanism that normally holds the patient chair. A transmission monitor was inserted prior to the collimator. We tried to minimize the fluctuations on the ratio of dose at the measurement point to the monitor signal by performing regular checks of and, if necessary, adjustments to the homogeneity of the lateral profiles. Nevertheless, on some occasions the adjustments induced differences of about 1% resulting in a significant contribution to the type A uncertainty.

Depth dose curves measured with a diode are shown in figure A.1 and lateral profiles obtained in air from diode measurements at the isocentre along the horizontal and vertical axis are shown in figure A.2. Correction factors for the lateral profile non-homogeneity, k_{profile} , were calculated by integrating two-dimensional lateral profiles over the projected area of the ICs perpendicular to the beam and correcting this to a similar integrated value for the SPGC core. The two-dimensional lateral profiles were measured using radiographic film.

Two sessions of measurements were performed, one in June 2003 and one in November 2003. In the modulated beam a dose of about 3 Gy was given in 12 s (0.25 Gy s^{-1}). For the non-modulated beam this was about 4 Gy in 15 s (0.27 Gy s^{-1}).

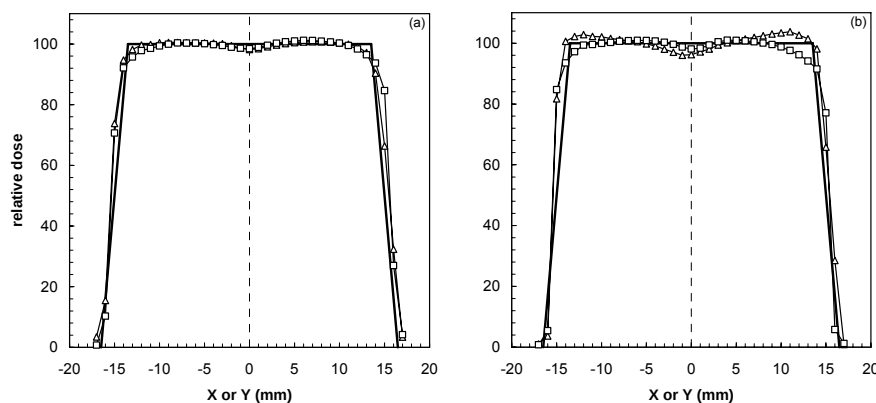


Figure A.2. Lateral profiles in the horizontal (squares) and vertical direction (triangles), measured with a diode for (a) the modulated beam and (b) the non-modulated beam. The thick curves are the profiles used in the heat transfer simulations.

A.3.2. Portable graphite calorimeter

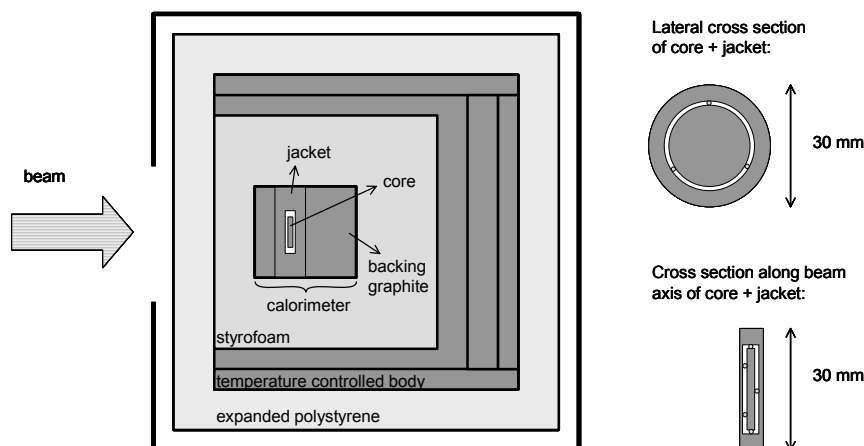


Figure A.3. Schematic diagram of the small-body portable graphite calorimeter (left, not to scale, adapted from McEwen and Duane, 2000) and cross sections of the core and the jacket (right, approximately to scale).

The SPGC used in this work is derived from NPL's previously existing PGC, which is described in detail by McEwen and Duane (2000) and McEwen (2002). The problem with the existing PGC was that due to the lateral size of the proton beam, only a small fraction of the jacket (the graphite surrounding the core) was irradiated, causing dramatic heat flows that rapidly affected the temperature drifts at the level of the core. For this reason, a SPGC was built such that the entire jacket was exposed to the beam. The materials were the same as those used by McEwen (2002), except that the expanded polystyrene surrounding the calorimeter was substituted by styrofoam. A schematic diagram of the SPGC is shown in figure A.3. The cylindrical core has the same dimensions as in the existing PGC: 20 mm diameter and 2 mm length. Four thermistors are inserted in the sidewalls of the core and pair wise coupled into opposing arms of two DC bridges. The thermistor material constitutes

0.01% of the total mass in the core, whereas the epoxy surrounding the thermistor beads contributes to 0.25% of the total mass. Using small, expanded polystyrene beads, the core is kept in place at the centre of a 22 mm diameter, 4 mm long air cavity. The jacket has an outer diameter of 30 mm as opposed to the 90 mm side-to-side dimensions of the octagonal slabs of the original PGC (McEwen and Duane, 2000). The graphite slab in front of the core has a thickness of 0.75 mm and the one beyond the core a thickness of 1.0 mm. For the modulated beam, an additional 7.75 mm of graphite slabs was added in front and 2.75 mm of graphite behind the core. Together with the 2.0 mm water equivalent thickness¹ of the styrofoam in front of the core, this resulted in the centre of the core being at a water equivalent depth of 17.0 mm. The amount of backing graphite was chosen such that the entire graphite body was in the homogeneous part of the SOBP. For the non-modulated beam, no additional graphite was placed in front resulting in a water equivalent depth of 4.8 mm for the centre of the core. 25.75 mm of graphite was added as backing material, an amount that was chosen such that the average energy deposition per unit mass in the backing body was estimated to equal the dose in the core in order to minimise the amount of heat transfer.

The measurements were analysed as described by McEwen (2002) by evaluating the calorimeter signal at mid-run, which compensates for the heat losses during irradiation (Seuntjens and DuSautoy, 2004). For each measurement, a linear fit to the pre-irradiation drift was extrapolated to mid-run. This was accepted after verifying that the use of polynomial fittings resulted in differences smaller than 0.05% of the calorimeter signal. Fourth order polynomials were fitted to the post irradiation drift curve for the modulated beam data, whereas sixth order polynomial fits were deemed necessary for the non-modulated beams since the curvature of the drift curve was more pronounced (see below). These polynomials were linearly extrapolated to mid-run by calculating the tangential at the end of irradiation. The temperature difference at mid-run was converted into absorbed dose to graphite by multiplying with the specific heat capacity, which was measured by Williams et al. (1993). In order to convert this into absorbed dose to water, the ratio of Spencer-Attix stopping power ratios water to air and graphite to air, $s_{w,air}^{\Delta} / s_{gr,air}^{\Delta}$, with an energy cut-off $\Delta = 10$ keV was calculated as described by Palmans and Verhaegen (2002) based on the ICRU 49 stopping power data. This was calculated at the effective energy, which is the mono-energetic proton energy for which the csda range equals the residual range R_{res} ², yielding a value of 1.119 for both the modulated beam ($R_{res} = 1.5$ cm) and the non-modulated beam ($R_{res} = 2.7$ cm).

A.3.3. Ionisation chambers

One cylindrical IC (NE2561 serial number (SN) 289 in the first session and NE2611 SN 193 in the second session) and two plane-parallel ICs (NACP02 SN 6306 and PTW Markus SN 2225) were used to compare with the SPGC. All ICs were calibrated in terms of absorbed dose to water in a ⁶⁰Co beam at NPL. The plane parallel ICs had in addition a calibration factor in NPL's 19 MeV electron beam. The IC measurements were performed in a separate graphite phantom. For various practical reasons (limitation of the workspace, thickness of graphite and styrofoam slabs available, density variations of the graphite slabs, etc.) it was not possible to position the ICs at exactly the same depth and distance as the SPGC core, as

¹ The water equivalent thickness of a slab is obtained by converting its thickness with the ratio of csda (continuous slowing down approximation) ranges of 60 MeV protons in water and in the slab material.

² R_{res} , expressed in cm, as defined in IAEA TRS-398 is the distance from the reference point to the depth where the dose is reduced to 10% of its maximum value.

can be seen in figure A.1. Any difference in the effective source to detector distance (SDD) and in the water equivalent depth of the IC compared to the SPGC was accounted for by correction factors k_{SDD} and k_{pdd} . The distance of the isocentre to the effective source was derived from a series of Markus chamber measurements at various distances and found to be 1.86 m. The factors k_{pdd} were determined from the percent depth dose (pdd) curves in figure A.1.

For the modulated beam a contribution to the positioning uncertainty is associated with the dose fluctuations as a function of depth due to the modulation technique (visible in figure A.1). These have a standard deviation of 0.3% and their effect will be pronounced for the plane-parallel ICs but much less for the cylindrical IC (where there is an averaging in depth due to the shape of the IC). A 0.3% contribution for this effect has been accounted for in the IC positioning uncertainty.

We have to note that for the cylindrical ICs in the modulated beam the concept of effective depth of measurement was used. For the NE2561/NE2611 this is, with respect to the centre of the IC, 2.9 mm closer to the beam source (Palmans and Verhaegen, 2000 and Palmans, 2000). For the plane-parallel ICs, the reference point was positioned as close as possible to the reference depth of 9.5 mm graphite in the modulated beam and 1.75 mm graphite in the non-modulated beam. The 0.6 mm graphite window of the NACP02 was accounted for, whereas for the Markus, which was used without its water-proofing cap, no window thickness was taken into account.

Absorbed dose to water was derived by applying IAEA TRS-398 (Andreo et al. 2000). The general formalism for absorbed dose to water $D_{\text{w},Q}$ in the proton beam quality Q is:

$$D_{\text{w},Q} = M_Q \cdot N_{D,\text{w},Q_0} \cdot k_{Q,Q_0} \quad (\text{A.1})$$

where M_Q is the IC reading corrected for influence quantities, N_{D,w,Q_0} is the calibration factor in the calibration beam quality Q_0 and k_{Q,Q_0} is the beam quality correction factor. For the results based on the ^{60}Co calibration factors, k_{Q,Q_0} (which is then simply denoted k_Q) is given in TRS-398 as a function of the beam quality $Q = R_{\text{res}}$, the residual range in cm.

However, for the results based on the electron calibration factors, the beam quality correction factor needed to be calculated. Referring to the electron calibration quality as Q_{e19} and introducing ^{60}Co as an intermediate beam quality³, $k_{Q,Q_{\text{e19}}}$ was calculated as:

$$k_{Q,Q_{\text{e19}}} = \frac{k_Q}{k_{Q_{\text{e19}}}} \quad (\text{A.2})$$

No recombination correction was applied to the IC measurements. Assuming the beam to be continuous - the pulse repetition rate is of the order 25.8 MHz – a theoretical calculation gives an estimated recombination correction of 0.1% for both the cylindrical and plane-parallel ICs, biased at 100V and 200V respectively. This is consistent with preliminary

³ Note that this intermediate beam quality has nothing to do with any experimental condition; see the cross calibration method for electron beams in IAEA TRS-398 (Andreo et al. 2000) for more details.

measurements at a four times higher dose rate that indicate a recombination correction of 0.3%.

A.3.4. Heat transfer simulations

Heat transfer simulations were performed using FEMLAB (Comsol 2003) using the thermal data that are listed in table A.1. For air they were taken from Kaye and Laby (1973), for styrofoam from the manufacturer's datasheets and for graphite they were all determined at NPL; the specific heat capacity, c_{gr} , by Williams et al. (1993), the thermal conductivity, k_{gr} , by McEwen and Duane (2000) and the mass density ρ_{gr} from weight and volume measurements.

Material	ρ (kg m ⁻³)	c (J kg ⁻¹ K ⁻¹)	k (W m ⁻¹ K ⁻¹)	$s_{\text{mat,gr}}$
Graphite	1767	726	134	1.000
Styrofoam	39.2	1400	0.027	1.102
Air	1.205	1006	0.0257	0.992

Table A.1. Values of the specific heat capacity, c , the thermal conductivity, k , the mass density ρ and the material (mat) to graphite stopping power ratios $s_{\text{mat,gr}}$ for the materials present in the SPGC used in the heat transfer simulations.

The geometry closely represented the real dimensions of the set-up. Between the jacket and the other plates, an air gap of 0.2 mm was modelled to account for the gap due to small pieces of tape keeping the front and back plates and the ring of the jacket together. For the gaps between the other plates no air gaps were modelled. Contrary to the findings of McEwen and Duane (2000), such air gaps resulted in complex post-irradiation drift behaviour in the simulations, which was not seen in the measured drift curves. This is likely due to a better thermal contact between the plates than in the original PGC. For the modulated beam, a homogeneous dose has been assumed over a 3.2 cm water equivalent range. For the non-modulated beam, the measured depth dose distribution of figure A.1 was taken into account. Doses in the three materials were scaled with ICRU 49 stopping power ratios given in table A.1, taking graphite as the reference material. The divergence of the beam and the lateral profiles were taken into account. The field width at the graphite phantom surface was thus 29 mm and the width of the penumbrae (90% to 10%) was 3 mm.

A.3.5. Monte Carlo simulations

A.3.5.1. Fluence correction factors

Fluence correction factors should be taken into account when converting doses from one phantom material to another (Palmans et al., 2002a and Schneider et al., 2002). This will be important for the conversion from dose in a graphite phantom to dose in a water phantom. In the present work, however, only a small correction factor, $k_{\Delta\text{fluence}}$, will be necessary to account for the effect of the change of the fluence correction factors from the depth of measurement in the calorimeter to the depth of measurement of the ionisation chambers. A modified version of the proton Monte Carlo (MC) code PTRAN (Berger, 1993), described

in detail by Palmans and Verhaegen (1997, 1998) was used. Stopping power data were taken from ICRU report 49 (1993) and non-elastic nuclear interaction data were taken from ICRU report 63 (2000).

A.3.5.2. Gap and volume averaging correction factors in the SPGC

Gap and volume averaging correction factors in the SPGC were calculated using another modified version of PTRAN that allows the simulation of cavities (Palmans et al. 2001, 2002b). The presence of the air gap between the jacket and the core is accounted for by a compensated gap correction factor, k_{gap} . This has been calculated as the ratio of the average dose to the core, calculated when it is sitting in the front of the gap which is filled up with graphite, to the average dose over the core when it is positioned in the middle of the air gap as is the case in the real measuring conditions. The volume averaging correction factor, k_{vol} , accounting for the finite dimensions of the core, has been calculated as the ratio of the dose in a small sphere at the centre of the core and the average dose over the core volume. A more detailed discussion of these procedures has been given by Palmans et al. (2004).

A.4. Results and discussion

A.4.1. Correction factors for the portable graphite calorimeter

A.4.1.1. Heat transfer

Figure A.4 compares the measured and simulated time dependence of the core temperature before, during and after the irradiation. For both the modulated and the non-modulated beams, the curves representing the measurements are the average of all the runs recorded in the proton beam after subtracting the linear pre-irradiation drift. The simulated temperature function was re-normalized at a time point three seconds after the end of the irradiation.

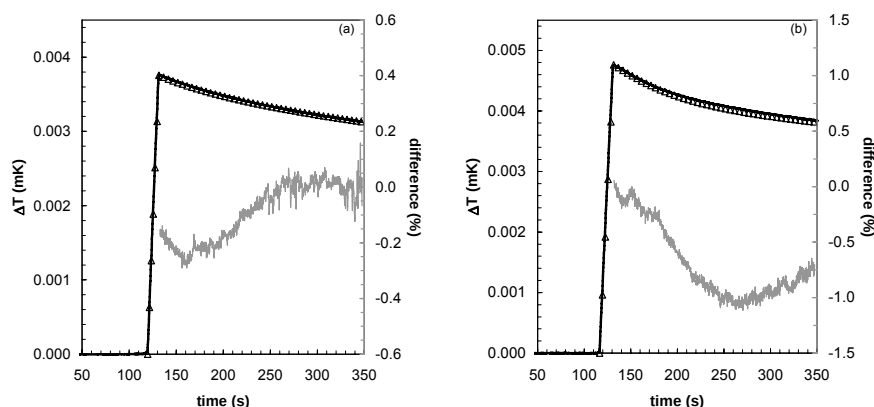


Figure A.4. Measured (black curves) and calculated (symbols) time dependence of the temperature in the SPGC core as a function of time during and after the irradiation (a) for the modulated beam and (b) for the non-modulated beam. The grey curves (scales on the right hand side) are the differences in percent between the measured and calculated data.

For the modulated beam there was a good agreement. Changes to the geometry, such as omitting the air gaps between the jacket and the build-up and backing plates, introducing the polystyrene foam beads between core and jacket or leaving air gaps between the SPGC and styrofoam resulted in very slight changes to the post irradiation drift curve. Linear extrapolation to mid-run of a fourth order polynomial fit to the simulated drift curve, in the same way as was done for the measurements, resulted in a correction factor of $k_{\text{extra}} = 1.001 \pm 0.001$. This correction was found not to depend significantly on the various geometrical changes mentioned above.

Figure A.4b shows that for the non-modulated beam the curvature of the post-irradiation drift curve is larger than for the modulated beam. The difference with the measurements is also larger. A better agreement could be obtained by increasing the field width by 0.5 mm. The reason for this could possibly be found in the inaccurate modelling of the lateral dose distribution in the plateau. No attempt was made to investigate this further, since the effect on the extrapolations to mid-run was, again, found to be small. Linear extrapolation to mid-run of a sixth order polynomial fit to the simulated drift curve, in the same way as was done for the measurements, resulted in a correction factor of $k_{\text{extra}} = 0.999 \pm 0.001$. When a fourth order polynomial fit was used, the correction factor amounted to $k_{\text{extra}} = 0.994$. This difference between the fourth and sixth order polynomials was the same as found in the experimental extrapolations, strengthening the confidence in the simulated correction factors.

A.4.1.2. Impurity correction

Another correction factor related to thermal phenomena is the impurity correction due to the different thermal properties of the thermistor material and surrounding epoxy compared to graphite. The thermal properties for the thermistor material are not known though the result is not very sensitive to these data given the small contribution to the total mass. Experimental data from Krauss (2002) for a typical thermistor material have been used. The data for epoxy, on the other hand, contribute to a larger extent and there is also considerable spread on the data, especially the specific heat capacity for this material for which values between 1400 and 2000 J kg⁻¹ K⁻¹ were found. Taking the heat capacity of these non-graphite components into account, the impurity correction varies between 0.14% and 0.27% depending on the data values assumed. A correction factor $k_{\text{impurity}} = 1.0020$ has therefore been taken into account with a standard uncertainty of 0.05%.

A.4.1.3. Gap and volume averaging correction factors

k_{gap} and k_{vol} at the reference points in both the modulated and the non-modulated beam were found to be unity within the type A standard uncertainties of the MC simulations (0.05%) indicating that proton scattering does not play a significant role in these corrections.

A.4.2. Correction factors for the ionisation chambers

A.4.2.1. Beam geometry related correction factors

k_{SDD} , k_{pdd} and k_{profile} calculated as described in section A.2, are given in table A.2. It is clear that some of these correction factors are substantial. Problematic for k_{profile} in the non-modulated beam was that, especially for the Markus, the result was very sensitive to the lateral position of the IC with respect to the centre of the profile. The magnitude of these variations was estimated by displacing the lateral profile over 1 mm in all directions. Based

on this, a standard uncertainty of 0.3% was estimated on k_{profile} for all situations, except for the Markus in the non-modulated beam where it amounted to 0.5%.

CF	Modulated beam			Non-modulated beam	
	NE2561	NACP02	Markus	NACP02	Markus
k_Q	1.038	0.989	1.003	0.988	1.003
$k_{Q,Q_{e19}}$	n.a.	1.113	1.119	1.112	1.119
k_{SDD}	1.007 / 0.999	1.006 / 0.999	n.a. / 0.999	1.007 / 0.999	1.007 / 0.999
k_{pdd}	0.999	0.999	0.999	1.011 / 1.014	1.010 / 1.014
$k_{\Delta\text{fluence}}$	0.999	0.999	0.999	0.999	0.999
k_{profile}	1.000	1.000	1.010	1.013	1.027

Table A.2. Correction factors (CF) for the ionisation chambers used in this study. The definition of the various correction factors is given in the text. In those cells where there are two values, the left one refers to the measurements in June 2003, the right hand one to those in November 2003.

A.4.2.2. Fluence correction factors

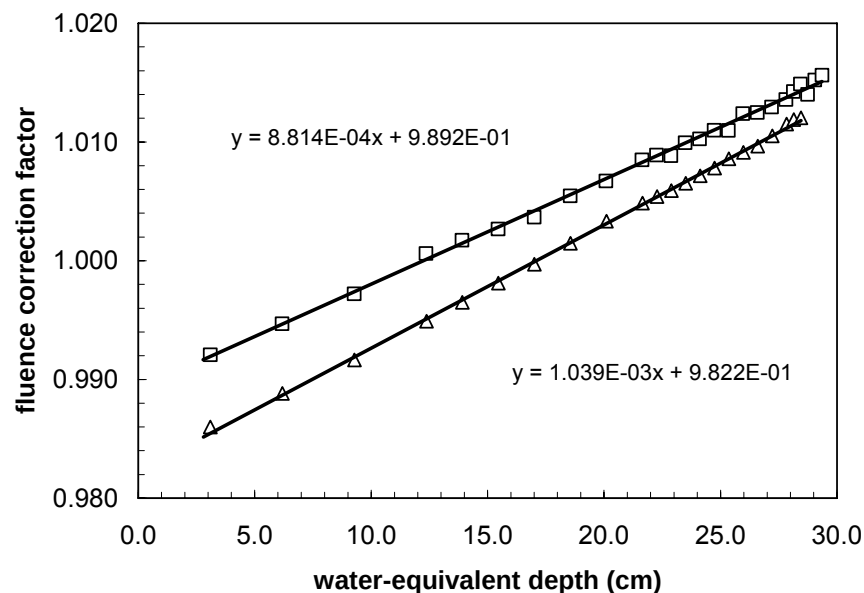


Figure A.5. Monte Carlo calculated fluence correction factors to convert absorbed dose to water in a graphite phantom to absorbed dose to water in a water phantom at equivalent depths for a modulated beam (squares) and a non-modulated beam (triangles).

Fluence correction factors for the conversion of dose to water in a graphite phantom to dose to water in a water phantom are shown in figure A.5. The observation in figure A.5 that the fluence correction factor does not equal unity at the surface, as could be expected, has been explained in detail by Palmans et al (2002a, 2004) and is due to the incorrect conversion of the contribution to absorbed dose from non-elastic nuclear interactions with the stopping power ratios. Palmans et al. (2004) showed that results obtained with different MC codes can vary substantially, but that the slope, which is of importance here, is very similar for all the MC codes used. The correction factors $k_{\Delta\text{fluence}}$, are listed in table A.2.

A.4.3. Comparison of the portable graphite calorimeter with ionisation chambers

Figure A.6 shows $D_{w,\text{SPGC}}/D_{w,\text{IC}}$, the ratio of absorbed dose to water obtained from the SPGC measurements and from the IC measurements after all correction factors discussed above were applied. Each data point represents a set of measurements which usually consisted of eight calorimeter runs with IC measurements prior to and after the set of calorimeter measurements. The standard deviation on a set of SPGC measurements was typically 0.6%. The uncertainty bars represent combined type A standard uncertainties for the calorimeter and IC measurements. The horizontal dashed lines represent the weighted mean for each IC type, beam quality and calibration modality.

No significant difference was observed between the measurements performed in June 2003 and those performed in November 2003. The different IC types were found to agree well within one calibration modality and beam type. There was, however, a systematic difference between the data obtained with ^{60}Co calibration factors and those obtained with electron calibration factors. This confirms earlier results reported by Thomas et al. (2004) and indicates that the ratio of the calibration factors and the ratio of the IC data from IAEA TRS-398 for ^{60}Co and electron beams are inconsistent. The issue was also raised by Andreo et al. (2002) and is presently under further investigation at NPL.

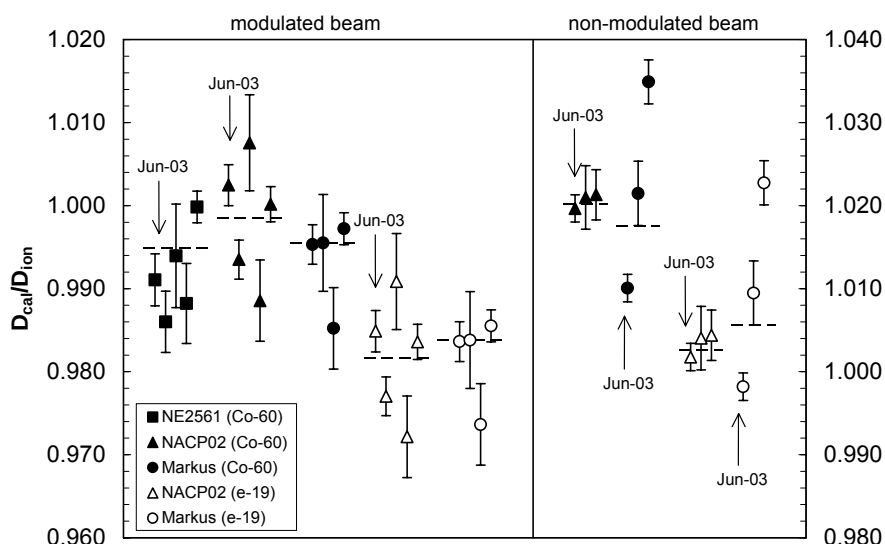


Figure A.6. Ratio of dose to water obtained with the SPGC and dose to water obtained with the ICs applying IAEA TRS-398. The data points measured in June 2003 are indicated.

For the modulated beam, the average $D_{w,SPGC}/D_{w,IC}$ ratio was 0.996 using ^{60}Co calibration factors and 0.983 using electron calibration factors. For the non-modulated beam it was 1.019 using ^{60}Co calibration factors and 1.004 using electron calibration factors. By accepting the data related to the calibration beams we are in principle only able to derive values for the product of the average energy required to produce an ion pair in dry air by protons, $(W_{\text{air}}/e)_p$, the water to air stopping power ratio in the proton beam, $(s_{w,\text{air}})_p$ and the overall IC perturbation factor in the proton beam, p_p . Based on the $(W_{\text{air}}/e)_p$ value of 34.23 J/C recommended in IAEA TRS-398, we could derive $(W_{\text{air}}/e)_p$ by accepting all the other data in IAEA TRS-398 to find values of 34.1 J/C using ^{60}Co calibration factors and 33.6 J/C using electron calibration factors in the modulated beam and 34.9 J/C using ^{60}Co calibration factors and 34.4 J/C using electron calibration factors in the non-modulated beam. The uncertainties on these values are discussed in the next section.

A.4.4. Uncertainties

A detailed uncertainty analysis was performed but is not discussed here in full, mainly because the discussion can be restricted to a few relatively large uncertainties that dominate the overall uncertainty budget. On the SPGC side, the uncertainty is dominated by the uncertainty on $s_{w,\text{air}}^\Delta / s_{\text{gr},\text{air}}^\Delta$. A standard uncertainty of 1% was taken from IAEA TRS-398 for proton stopping power ratios in the clinical energy range, which is consistent with the lower values quoted in ICRU report 49. For the ICs, the uncertainty on both the ^{60}Co as the electron beam calibration factors is 0.75%. The standard uncertainties on the k_Q values are 1.7% for the cylindrical IC and 2.1% for the plane parallel ICs. The standard uncertainty on the $k_{Q,Q_{e19}}$ values for the plane-parallel ICs was obtained by combining the uncertainties for the proton beam data and the electron beam data in IAEA TRS-398 and was found to be 1.4%. Other substantial contributions were 0.2% for the reproducibility of the PGC - IC response ratio (0.4% for the Markus in the non-modulated beam) and 0.3% for the lateral profile correction factors (0.5% for the Markus in the non-modulated beam). This results in standard uncertainties on the ratio $D_{w,SPGC}/D_{w,IC}$ of 2.1% for the NE2561 and 2.5% for the plane-parallel ICs using ^{60}Co calibration factors, and 1.9% for the plane-parallel ICs using electron calibration factors. These numbers are also the standard uncertainties with which we are able to derive $(W_{\text{air}}/e)_p$. Leaving out the uncertainties on the beam quality correction factors yields 1.3% to 1.4%, which is thus the standard uncertainty with which we were able to determine k_Q or $k_{Q,Q_{e19}}$ values.

A.5. Conclusions

In this work we described the construction and performance of a dedicated small-body portable graphite calorimeter for low-energy clinical proton beams. The standard deviation for 12s runs at a dose rate of 15 Gy min⁻¹ was typically 0.6%. Perturbation correction factors for the gap effect, volume averaging effect, heat transfer phenomena and the impurity effect were calculated and applied in a comparison experiment with IC dosimetry using the procedures and data from IAEA TRS-398. For the ionisation chambers a number of substantial correction factors were due for beam profiles and the geometrical set-up.

The $D_{w,SPGC}/D_{w,IC}$ ratio varied between 0.983 and 1.019 depending on the beam type and IC calibration modality. Given the uncertainties, which varied between 1.9% and 2.5%, this is not significantly different from unity, though the uncertainties are dominated by contributions from the calibration factors, beam quality correction factors and water to graphite stopping power ratios. The $(W_{\text{air}}/e)_p$ values inferred from the measurements varied

between 33.6 J/C and 34.9 J/C, with uncertainties that varied between 1.9% and 2.5% depending on the beam type and IC calibration modality.

The following improvements for the PGC and the measurement set-up in general will enable a substantial reduction of the overall uncertainty:

- reduction of the size of the temperature controlled body will establish quicker settling and will make measurements more efficient by allowing shorter time intervals; a smaller calorimeter will also allow the use of an external monitor between the final collimator and the SPGC or IC phantom in the CCO beam
- IC measurements should be performed in a phantom that allows accurate positioning at the same depth as in the SPGC (especially for non-modulated beams),
- making the core size to match the IC size will allow minimising the uncertainty due to lateral beam non-uniformity,
- removing the copper positioning cross wires, which are the main source of the lateral profile non-uniformity, for precision dosimetry work
- calibrations of the ICs in terms of absorbed dose to graphite directly to the SPGC and a detailed analysis of the uncertainties on all remaining data needed could reduce the uncertainty on the determination of $(W_{\text{air}}/e)_p$.

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ADDENDUM B: PALMANS (2003)

PALMANS, H. Effect of alanine energy response and phantom material on depth dose measurements in ocular proton beams. *Technol. Cancer Res. Treat.*, 2003, **2**, 579-586.

B.1. Abstract

The usefulness of the alanine dosimeter for proton beams has been demonstrated before. In this paper a literature overview is given of the application of alanine in proton beam dosimetry. The energy dependence of the dosimeter has been investigated and Monte Carlo calculations were performed to investigate the influence of this energy dependence on a Bragg curve as well as to explain some phenomena related to the use of alanine in a solid phantom that were reported in the literature.

B.2. Introduction

The treatment of eye melanoma is up to now the most important application of proton therapy. For the treatment of these lesions ocular proton beams are used, which are low-energy beams with a range smaller than 3.5 cm in water corresponding with energies below 65 MeV. This treatment technique takes advantage of the depth dose distribution of a proton beam exhibiting a high dose peak at the end of the proton range, the Bragg peak, beyond which the dose falls rapidly to zero.

Crystalline alanine has been used since the 1960's for dosimetry of light and heavy ions and has gained renewed attention as a potential absolute dosimeter recently. Usually, alanine is used in powder form and mixed with a binding agent such as paraffin or cellulose in order to shape it into small dosimeters that can be easily handled. Alanine dosimetry is based on the measurement of the amount of radiation-induced radicals formed by means of the electron spin resonance (ESR) technique. By applying a magnetic field and then supplying electromagnetic energy in the microwave frequency range, magnetic dipole transitions are induced to the spin states of the unpaired electron in the radical. Quantification of the absorbed microwave energy around the resonance frequency yields the ESR spectrum, the amplitude of which is a measure of the amount of radicals formed. Advantages of the alanine dosimeter are that the signal is preserved after read-out and exhibits a long time stability, the composition is nearly tissue equivalent and the dose response is linear over a wide dose range. Disadvantages are that the sensitivity of the technique is rather low such that the required doses are an order of magnitude larger than usually applied in a radiotherapy treatment fraction and that the ESR technique is quite elaborate.

Bradshaw et al. (1) and Ebert et al. (2) compared the response of alanine powder in proton beams to the response in ^{60}Co gamma rays and later, similar experiments were performed with pellets of alanine powder in a cellulose matrix by Hansen and Olsen (3). More recently, Fattibene et al. (4) and Onori et al. (5,6) have performed a considerable amount of work using alanine pellets calibrated in a ^{60}Co beam. They determined the dose in the centre of the spread-out Bragg peak⁴ (SOBP) of a 60 MeV modulated proton beam (effective energy ± 45 MeV) and in the entrance region of a non-modulated beam (effective energy ± 60 MeV) and compared it with the response of a Markus-type plane-parallel ionisation

⁴ An SOBP is obtained by smearing out the Bragg peak along the beam direction such that a flat dose as a function of depth is achieved. This can be done by changing the energy, and correspondingly the range, of the beam as a function of time, or by a range modulating wheel with a variable thickness.

chamber. They systematically found a 3% lower dose with their alanine system, which is within the standard uncertainty of the ratio that was estimated to be 6% containing a contribution of 4% on the average energy required to produce an ion pair in dry air, $(w_{\text{air}})_p$. They thus derived a relative effectiveness (RE) of alanine of 0.97. RE in this sense is defined as the ESR signal of the alanine detector to a certain proton dose divided by the ESR signal due to the same dose in a ^{60}Co beam. It is worth noting that for the ionisation chamber dosimetry, they used the $(w_{\text{air}})_p$ -value recommended by the European Charged Heavy Particle Dosimetry Group (ECHED) (7, 8), which is 3% higher than values resulting from comparison of ionisation chamber dosimetry with calorimetry (9). The $(w_{\text{air}})_p$ -value is an essential parameter in ionization chamber dosimetry since it converts charge measured in air into dose or a related quantity. They also noticed a slight decrease of the RE with decreasing proton energy. Cuttone et al. (10) performed a similar comparison, using a Markus plane-parallel and an Exradin T1 cylindrical ionisation chamber at effective energies of about 40 MeV and 60 MeV and found an agreement within 2%. Again this agreement is better than the quoted combined standard uncertainty of 5% on the ratio of both results. Bartolotta et al. (11) compared alanine with a Markus chambers at lower effective energies of 24 MeV and 14 MeV and found that the alanine dose-response was respectively 4% and 8% lower, indicating that the RE decreases significantly upstream of the Bragg peak of a low-energy proton beam. Recently, Fattibene et al. (12) did total absorption experiments in proton beams with energies of 6 MeV and lower and found a decrease in RE to 60%. Waligorski et al. (13) combined experimental data from several publications (1, 2, 3, 14, 15) on the variation of the RE with energy for protons and heavy ions and compared these data with results of calculations based on track structure theory. Their results indicate the same qualitative decrease of alanine response with decreasing energy (or increasing LET). The theoretical values they found are in general higher than the experimental values, but lower than the results of Bartolotta et al. (11). For higher atomic number ions the agreement between model and theory was in general better.

Alanine has also been investigated as a relative detector and several authors have investigated its suitability for depth dose measurements compared to other detectors (4, 5, 6, 11, 16, 17) in general with satisfying agreement with plane-parallel ionisation chambers. The advantage of this detector is that the stopping power over the entire clinical proton energy range is close to that of water (6) and whilst the density of pure alanine is quite high (1.42 g cm^{-3}) compared to water, the density of the mixtures in which it is in general used for practical purposes is much closer to water. The pellets in which it is often compressed together with paraffin wax, polyethylene or cellulose as a binding material have typical densities varying from 1.05 to 1.26 g cm^{-3} . Alanine-film, in which typically 30% by weight alanine grains are bound in polyethylene have relative densities around unity. Several authors confirmed the linearity of alanine pellets in the accumulated dose range of 5 to 300 Gy (6, 16, 17). A disadvantage is that the sensitivity of the technique is not particularly high. Pellets of 2 mm thickness and 5 mm diameter require doses of more than 10 Gy for a standard deviation on the read-out of 1%, whereas alanine films require over 100 Gy for reducing the standard deviation below 10% (6). For pellets, another general disadvantage observed and admitted by most of the quoted authors is the volume averaging effect in low-energy beams due to the relatively large size of the pellets. When measuring a depth dose curve using a stack of pellets in a solid phantom (PMMA or solid water), Onori et al. (6) observed dose tails beyond the Bragg peak, which they explained as being due to air gaps between the detectors and the surrounding material.

In the present paper, the alanine response as a function of energy according to the literature discussed above is investigated. The influence of this energy dependence on the measurement of Bragg curves is evaluated and discussed. To this end, Monte Carlo

calculations were performed using a modified version of PTRAN. The effect of placing a stack of pellets in a solid water phantom in order to measure a depth dose curve is studied as well.

B.3. Materials and methods

B.3.1. Data needed for Monte Carlo simulations

Proton stopping power data (the average energy loss of a charged particle per unit of trajectory) are required for performing a Monte Carlo simulation. For alanine, these data are not given in ICRU report 49 (18) and the most common approach in that situation is to derive them from the stopping powers for the elemental constituents using Bragg's additivity rule, which sums the stopping powers of the elementary constituents, weighed by their fractions by weight. This was for example done by Onori et al. (6). However, it is admitted in ICRU 49 that the mean excitation energy derived that way does not account for the chemical electron states in the molecular environment. Therefore, in the present work, the alanine stopping powers have been calculated using the mean excitation energy value of 71.9 eV, consistent with the procedures outlined in ICRU report 37 (19) to account for chemical binding effects, instead of the value of 69.3 eV that would result from Bragg's additivity rule.

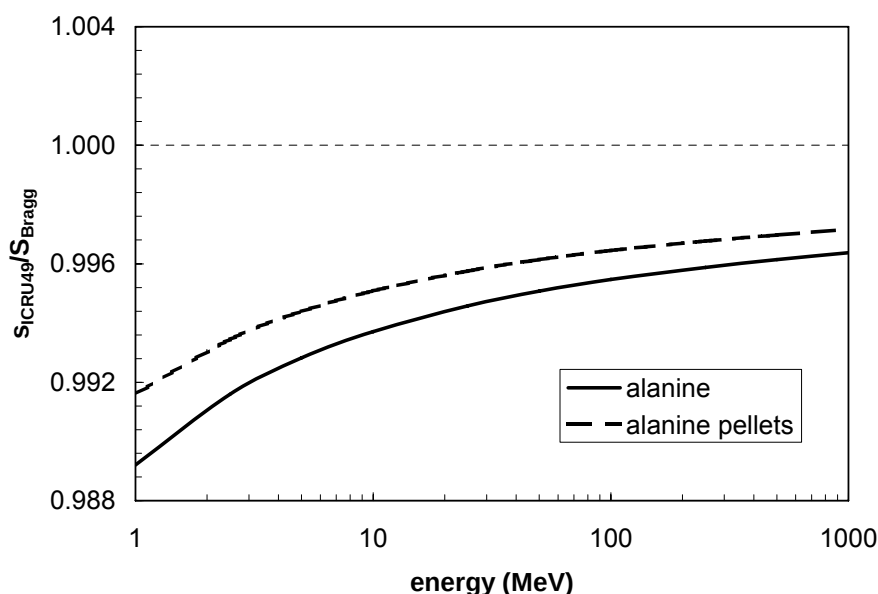


Figure B.1. Ratio of mass stopping powers calculated according to the recommendations in ICRU 49 (including consistent values for the mean excitation energy) and those calculated from the elemental constituents applying Bragg's additivity rule.

A second correction that is made for the present work is regarding the so called Barkas correction, which is also obtained incorrectly when applying Bragg's rule, since it does not account for the difference between molecular hydrogen and hydrogen in compounds. The Barkas correction accounts for the difference in stopping power for a particle and its anti-particle and was calculated in accordance with the recommendations of ICRU reports 37 and

49. The errors regarding the density correction made by applying Bragg's rule are considered not to affect the stopping powers in the clinical energy range.

Figure B.1 shows the ratio of stopping powers for alanine and for Onori's alanine pellets, consisting of 80% alanine and 20% paraffin by weight, in the clinical energy range calculated in this work to those obtained from Bragg's rule. It is obvious that at lower energies the differences are substantial and tend to compensate the increasing alanine to water stopping power ratio with decreasing energy, which was shown by Onori et al. (6). The variation of the water to alanine stopping power ratio from 1 to 250 MeV thus becomes less than 1% and from this, we can conclude that the stopping power ratio will not have a substantial influence on the shape of a depth dose curve.

Another data set needed for a Monte Carlo simulation are the non-elastic nuclear interaction cross section. The most consistent data are those from ICRU report 63 (20) although the stated standard uncertainties are considerable (5 to 10% for the total cross sections). Some recent papers indicate that the total interaction cross sections for low-Z materials of ICRU 63 are consistent with measured depth dose curves and thus these total interaction cross sections are used in the present work. The total nuclear interaction cross section per nucleon is calculated from the cross sections for elemental constituents by a weighted sum using the fractions by weight. In the modified version of PTRAN described below, the simplifying assumption is made that the fraction of the energy transfers in these nuclear interactions going to charged secondary particles is absorbed on the spot, whereas the rest going to neutrons and photons escapes from the geometry. This assumption has no significant effect in the energy range where the variation of the alanine response with energy is important.

B.3.2. Monte Carlo calculations

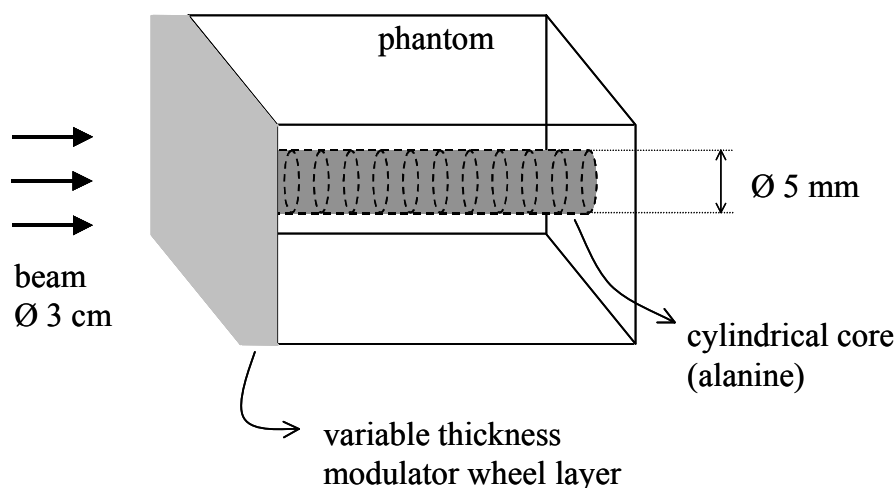


Figure B.2. Geometry for the Monte Carlo simulations.

The transport code used in this work is based on the proton Monte Carlo code PTRAN (21) that simulates mono-energetic pencil beams in homogeneous water without transport of secondary particles. Palmans and Verhaegen (22, 23) made adaptations in order to simulate transport in low-Z materials other than water, to allow the simulation of modulated proton beams, rectangular or circular beams and beams with a certain energy distribution. The implementation of a boundary-crossing algorithm allowed the simulation of inhomogeneous geometries (23). For the present work, a version of the code was used which has been

developed earlier for the simulation of the response of a stack of thermoluminescent dosimeters (TLD) in PMMA (24).

The simulated geometry is shown in figure B.2; a cylindrical rod of alanine or alanine pellet material is centred on the beam axis and surrounded by the phantom material. The scoring algorithm of PTRAN has been modified such that the dose at a chosen grid along the beam direction is scored in the region defined by the central rod only. In PTRAN, dose is scored at planes by multiplying the particle fluence with the electromagnetic and non-elastic nuclear stopping powers. For the latter contribution, only the fraction of the energy loss that is transferred to charged particles is taken into account and assumed to be deposited on the spot. The code has also been adapted to include an energy dependent detector material response. That way, a depth dose curve convolved with the energy response of a detector can be calculated. The effect of a polystyrene modulator wheel was simulated by adding an additional layer in front of the phantom, the thickness of which is sampled from the distribution of the thickness in the wheel for each individual incident particle.

Simulations were performed for circular beams with a diameter of 3 cm. The energies were 60 MeV mono-energetic and 62 MeV with a mono-energetic or gaussian energy distribution. The sensitive material in the central rod, with a diameter of 5 mm, was either alanine or Onori's alanine pellet composition of 80% alanine and 20% paraffin with a density of 1.26 g cm^{-3} , corresponding to the 1 mm pellets. The phantom material was either set to the same as the sensitive material or to water⁵. Depth dose curves were calculated with and without taking into account the energy response of the detector material.

B.4. Results and discussion

B.4.1. Discussion of the literature data

Table B.1 summarises all data from the references quoted in the introduction listed in chronological order. The experimental results can essentially be divided into two groups. In the first, which is indicated in table B.1 by the method of 'total absorption', a beam of low-energy protons is completely absorbed in the alanine dosimeter and the total deposited energy is derived from a measurement of the number of incident protons. The energy absorbed per unit mass is then derived by accounting for the fraction of the detector's volume that has been exposed. In the second group, a thin absorbing alanine layer, with a thickness that is small compared to the range of the protons passing through is compared with another detector to measure the dose deposited in the volume of alanine. In both experiments, RE is obtained by taking the ratio of the ESR signals per unit absorbed dose or unit of energy absorbed per unit mass in the proton beam and in a ^{60}Co beam. The effective energy E_{eff} given in table B.1 can be considered as a measure for the average LET of the protons in alanine. For the total absorption experiments, the average LET was calculated as the incident energy, divided by the range in the continuous slowing down approximation (csda)⁶ and the energy of mono-energetic protons having the same LET was taken as effective energy. For the other experiments, where only a fraction of the total proton energy was deposited in the alanine detector, E_{eff} was estimated as the average of the entrance and exit energies. It is also noticable, that there were a few experiments done in the sixties and one in the eighties and that there has been a renewed and intensive interest in applying alanine to protons in the last five years, which can be explained by the recent increase of the

⁵ Water was used as a replacement of solid water type WT1, which was used by Onori et al (6). WT1 is water equivalent in photon beams, but its equivalence to water in proton beams has not yet been investigated.

⁶ The csda range is the distance an individual proton would travel when it would loose energy at a rate equal to the average energy loss for an infinite number of protons.

amount of radiotherapeutic treatments using protons together with the awareness that there is a need of bringing the accuracy of proton dosimetry to the same level as in high-energy x-rays and electron beams.

Reference	E_{eff} (MeV)	RE	u_C	Method or Ref. Instr.
Bradshaw et al. (1)	6.8	0.690	0.140	Scintillator
Ebert et al. (2)	4.9	0.640	0.130	Total absorption
Ebert et al. (2)	6.1	0.610	0.120	Total absorption
Ebert et al. (2)	7.0	0.690	0.140	Total absorption
Ebert et al. (2)	7.7	0.700	0.140	Total absorption
Hansen and Olsen (3)	12.3	1.000	0.110	Total absorption
Hansen and Olsen (3)	2.9	0.860	0.030	Partial absorption
Onori et al. (6)	45.0	0.970	0.058	Ion chamber
Onori et al. (6)	60.0	0.970	0.058	Ion chamber
Bartolotta et al. (11)	23.6	0.955	0.034	Ion chamber
Bartolotta et al. (11)	13.7	0.908	0.032	Ion chamber
Cuttone et al. (10)	40.0	1.017	0.051	Ion chamber
Cuttone et al. (10)	60.0	0.990	0.050	Ion chamber
Fattibene et al. (12)	1.2	0.607	0.042	Total absorption
Fattibene et al. (12)	1.4	0.614	0.043	Total absorption
Fattibene et al. (12)	1.9	0.633	0.044	Total absorption
Fattibene et al. (12)	2.4	0.652	0.046	Total absorption
Fattibene et al. (12)	2.9	0.623	0.056	Total absorption

Table B.1. Relative effectiveness (RE) of the alanine dosimeter from the literature, with the uncertainties u_C as quoted in the references. The effective energy E_{eff} is a measure for the average LET of the proton spectrum in the detector. The fifth column indicates the method or the reference instrument with which alanine has been compared.

The data with reference to an ion chamber need some discussion since it is known that the $(w_{\text{air}})_p$ value varies considerably from one ion chamber dosimetry protocol to another and that an international consensus on one particular value has not as yet been established. The RE value of 0.97 found by Onori et al (6) applying the ECHED protocol (7, 8) would be 1.00 if the recommendations of IAEA TRS-398 (24) had been used. This difference is determined by the different values for the product of the $(w_{\text{air}})_p$ value and the water to air mass stopping power ratios that are recommended in both reports. Nevertheless, since the differences in the $(w_{\text{air}})_p$ values adopted in the various protocols are a reflection of the uncertainty with which it is known and these uncertainties are included in the uncertainty budgets quoted by the authors of the data in table B.1 (4, 5, 6, 10, 11), the data have been used in the remainder of this work as they were published with their given uncertainties. It

can be argued that this might induce inconsistencies in the data set, but the consistency of comparing data obtained from a comparison with an ion chamber to data obtained by other methods such as total absorption and scintillators suffers anyway from the lack of knowledge of the basic interaction data that come into the calculation of the delivered proton dose. As will become clear below, the overall picture of the energy dependence of the alanine response is dominated by other parameters than the $(w_{\text{air}})_p$.

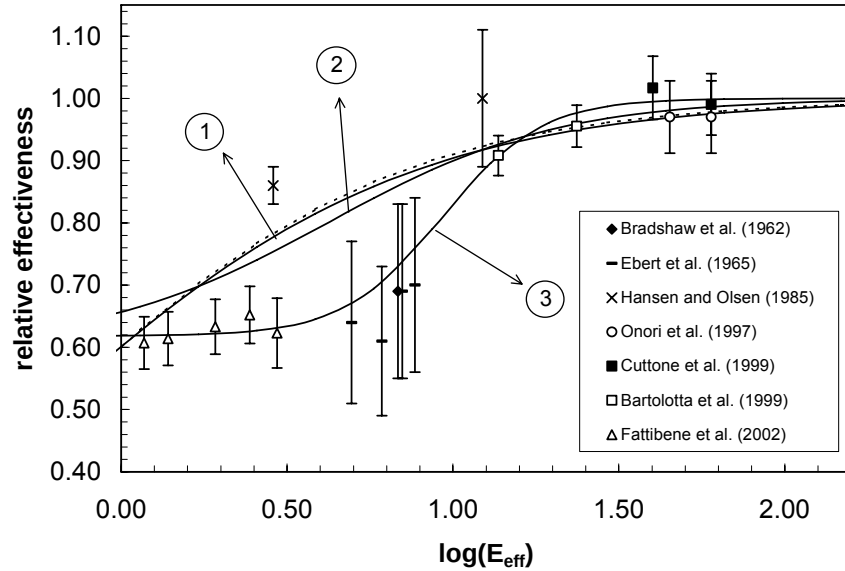


Figure B.3. Experimental values of the relative effectiveness of alanine as a function of effective proton energy, given by various sources in the literature. The thin lines numbered 1 to 3 are fits to the experimental data as discussed in the text.

In figure B.3, the data from table B.1 have been plotted as a function of the effective energy E_{eff} . It is noticeable that with some exceptions, there is a group of data at low-energy beams, mostly generated at Van de Graaff accelerators, for which RE obtained by total absorption experiments is concentrated around a value of 0.62 and a group of higher-energy beams, generated at cyclotrons for which the RE values are close to unity. This marked step in RE as a function of energy can not be directly explained by the track structure model of Waligorski (13) and it is not obvious to fit a function to these data. It has also been indicated previously (12) that inconsistencies might be due to differences in the alanine system and the ESR system used, so the uncertainty of such fits might be rather large and therefore, various fits have been tried out.

Since theoretical models (13, 14) predict saturation effects both at low and at high energies, a number of weighted sigmoidal fittings were tried, some of which are shown in figure B.3. The equation representing these curves is given by

$$\text{RE}(E_{\text{eff}}) = \text{RE}_0 + \frac{\Delta \text{RE}}{1 + e^{-C \cdot (\log(E_{\text{eff}}) - \log(E_m))}} \quad (\text{B.1})$$

In this equation, RE_0 , ΔRE , C and E_m are parameters and their values for the different fits have been given in table B.2. The curves in figure B.3 can be considered as extreme cases. The saturation level at high-energy was fixed to unity for all fits. For the curve indicated with number 1, the only additional condition was to set a lower limit of zero to the function. The dotted curve presents a similar fit indicated with 1' in table B.2, in which the data points with uncertainties larger than 10% were left out, to demonstrate that these points do not contribute substantially to the outcome. For the curve indicated with number 2, the lower limit of the sigmoid was set to 0.6, which is close to the bulk of the values at low-energies. Curve number 3 was obtained by ignoring the one outlier from Hansen and Olsen at $\log(E_{\text{eff}}) \approx 0.46$ and leaving the lower limit as a free parameter. The lower limit thus obtained was 0.618. The influence of these different fittings on calculated depth dose curves or on the conversion of measured depth dose curves is discussed in the next section.

Curve no	ΔRE	$\log(E_m)$	C	RE_0
1	1.0000	-0.2238	1.8322	0.0000
1'	1.0000	-0.2165	1.9095	0.0000
2	0.4000	0.6219	2.8921	0.6000
3	0.3819	0.9673	6.7193	0.6181

Table B.2. Parameters of the sigmoidal curves fitted to the data given in table B.1.

Strictly, in the total absorption experiments E_{eff} should be derived such that a mono-energetic proton with energy E_{eff} has the same response as the average over the total path of the proton. This has the disadvantage that the E_{eff} values for the different experiments depend on the response curve itself, such that the response curves should be determined iteratively. To have an idea how important this effect is, E_{eff} as a function of the incident energy, E_{in} , has been calculated such that the $RE(E_{\text{eff}})$ equals the integrated response $REI(E_{\text{in}})$, which is calculated using the continuous slowing down approximation as

$$REI(E_{\text{in}}) = \frac{\int_0^{z_{\text{max}}} RE(E(z)) \cdot \frac{dE}{dz} \cdot dz}{\int_0^{z_{\text{max}}} \frac{dE}{dz} \cdot dz} \quad (\text{B.2})$$

where dE/dz is the mass stopping power multiplied with the mass density and z_{max} is the maximal penetration of the beam. Figure B.4 shows the relation between E_{eff} and E_{in} derived from the three response curves of figure B.3, as well as the relation that uses the equivalent LET approach as described above. There are obvious differences but the curve obtained using the LET approach is not far from the average for the three other curves. Applying the effective energy values from figure B.4, curves 1 and 2 in figure B.3 would shift slightly towards lower energies, whereas curve 3 would shift to slightly higher energies. Since these differences will only affect the response for energies below 20 MeV, the effect will only be visible in the last few millimetres or thus in the Bragg peak of the proton beam, and will not

have a major quantitative effect. For the present work, the influence of the response on depth dose curves has been evaluated using the curves of figure B.3. It could be noted that qualitatively, curves 1 and 2 are in better agreement than response curve 3 with the theoretical considerations based on track structure theory by Waligorski et al (13), in the sense that the reduced response occurs at energies below a few MeV. Since the theoretical curve from Waligorski et al saturates at an unrealistic level of 0.75 at high energies, however, no attempt was made to include this response curve in the present work.

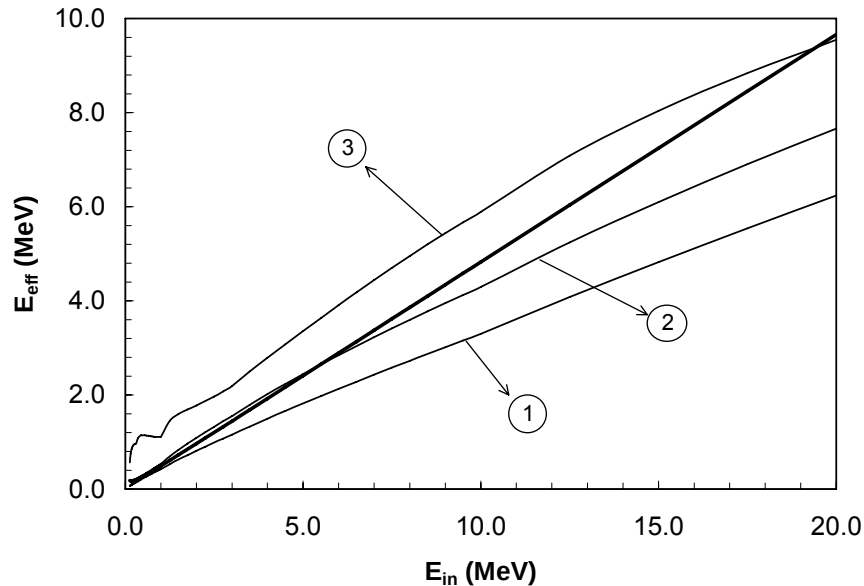


Figure B.4. Effective energy as a function of incident energy for the total absorption experiments. The thin lines are obtained such that the response in the total absorption of the energy E_{in} is the same as the response of the mono-energetic effective energy E_{eff} , according to response curves 1 to 3. The thick line is obtained such that the average LET in the total absorption experiment equals the LET at E_{eff} .

B.4.2. Influence of response curves and phantom material on Bragg peak

Figures B.5a and B.5b show calculated Bragg curves for a non-modulated and a range modulated 60 MeV mono-energetic proton beam. Note that for the non-modulated beam only a fraction of the depth dose curve is shown; at the phantom entrance all curves coincide. The sensitive and phantom materials were both set to alanine. The upper curve is the physical dose deposited in alanine, whereas the three other curves indicated with 1-3 were obtained by reducing the local energy deposition in the simulation according to the corresponding response curves in figure B.3.

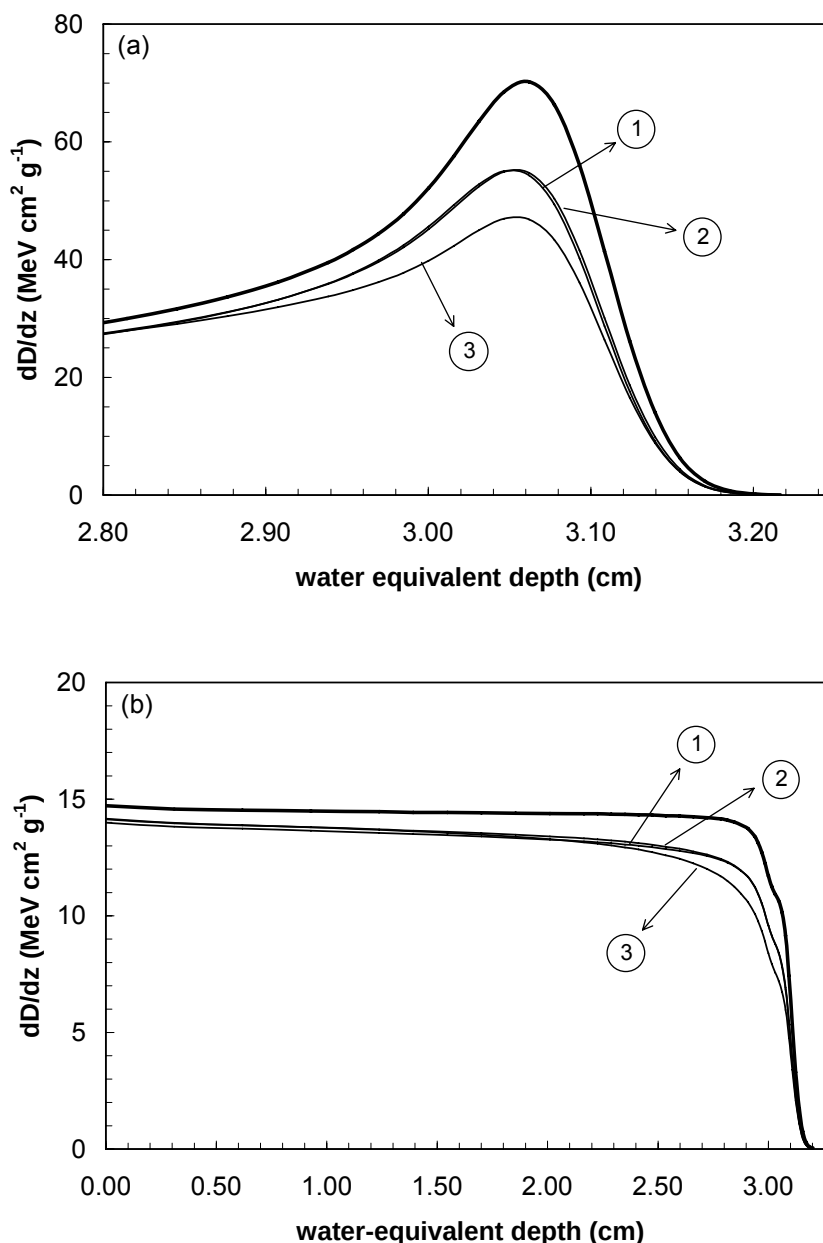


Figure B.5. Physical depth dose distributions in homogeneous alanine (thick solid curves) and response distributions taking into account the energy response functions derived from figure B.3 (thin solid curves), (a) for a non-modulated 60 MeV proton beam (only the distal fraction of the curve is shown) and (b) for a modulated 60 MeV beam.

For the non-modulated beam, it is obvious that whatever response curve is correct, a 20 to 30% reduction of the signal in the Bragg peak can be expected compared with a detector having a flat energy response. For the modulated beam this results in a reduction of about 5% over most of the spread out Bragg peak and 10 to 15% near the distal edge. This is a direct consequence of the experimental data shown in figure B.3, which was based on information from the literature. On the other hand, it is contradictory to the conclusions of Gall et al. (17) and Onori (6) who both found a reduction of only about 5% in the Bragg peak of a non-modulated beam, although Onori et al (6) admit that the uncertainty of their determination with alanine film is large (10%). For their results with alanine pellets it is

difficult to discriminate between the effect of volume averaging and the RE. Another point of concern is that in the frontal edge of the Bragg peak, Onori's results do not show the dose reduction either. This could be due to the use of a range scaling method based on electron densities rather than the csda range. Using a csda range scaling the results for alanine would shift to slightly larger depths, which would qualitatively correspond with the predicted reduction of the alanine signal, but would quantitatively still underpredict the effect. It is not clear what is the reason for these discrepancies between various sources from the literature. Although there could be a potential effect from the read-out procedure, the sources that discussed this aspect found that the ESR spectrum of alanine irradiated by protons was not different from that obtained by exposure to ^{60}Co gamma rays (3, 6, 16).

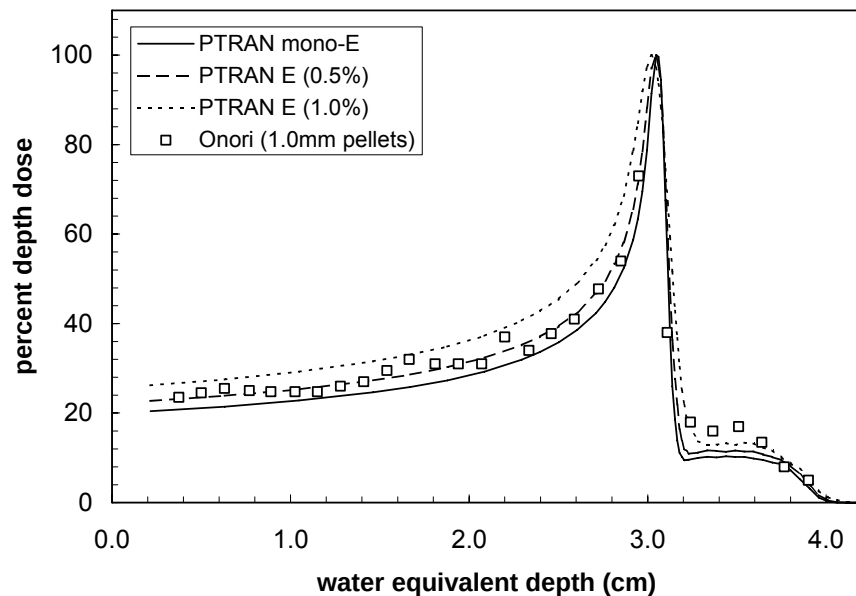


Figure B.6. Depth dose distributions measured with a stack of alanine pellets in solid water by Onori et al (6) (symbols) and calculated by Monte Carlo using gaussian energy distributions with a mean value of 62 MeV and variances of 0, 0.5 and 1% (curves).

The effect of a difference in the radiological properties between alanine and the surrounding phantom material is illustrated in figure B.6. The data points represent experimental results reported by Onori et al (6), which were normalised to their Markus ion chamber measurement. The curves in figure B.6 are all Monte Carlo calculated depth dose distributions in alanine pellet material (without taking an energy response effect into account) for various incident gaussian energy spectra. They were all normalised at the dose maximum to the same ion chamber depth dose curves and the depth was shifted such that the 80% dose levels at the distal edge of the Bragg peak coincided with the ion chamber result. This shift accounts for any unknown absorbers (air, scatter foils...) between the proton source and the phantom surface. The calculated depth dose curves were corrected for the inverse square law based on a source to surface distance of 100 cm, to account for the difference in the experimental conditions where a diverging beam is used and the situation in the Monte Carlo calculation where a parallel beam was simulated. The curve with a variance on the incident energy of 0.5% shows the best agreement with the experimental data points, but it is not sure whether or not this is the best match given the discussion in the previous paragraph. It is clear from figure B.6 that the tail beyond the Bragg peak is for a

major part explained by the inscatter of protons from the surrounding phantom into the stack of pellets, since in the simulations this is the only potentially significant source of protons at depths larger than 1.04 times the csda range in the alanine pellets. In this sense (part of) the origin of the tail is similar to the phenomena described by Goitein and Sisterson (26) behind bone and Teflon inhomogeneities in water. As indicated by Onori et al (6), the remainder of the effect could probably be explained by an additional amount of protons that tunnel through the air gaps between the pellets and the plastic phantom.

B.5. Conclusions

In this paper, an overview of the literature regarding the use of alanine as a dosimeter in clinical proton beams was presented. It was found that water to alanine stopping power ratios and non-elastic nuclear cross section ratios have a minor influence on the shape of a depth dose curve. Energy response data were extracted from sources that compared the absolute response of alanine in protons and in ^{60}Co gamma rays and various energy response curves were fitted to those data. Monte Carlo calculations were performed to investigate the effect of these energy response curves on measured depth dose distributions as well as to study the effect of the surrounding phantom material. The results showed a significant reduction of the Bragg peak, both in a non-modulated and a modulated beam. The calculated reduction of the response, however, was found to be larger than observed by a number of other investigators who measured depth dose distributions with alanine. The Monte Carlo results further show that dose tails, which are sometimes observed beyond a measured Bragg peak, can be explained by inscatter from the surrounding phantom material when the range in the material is larger than in the alanine pellet material. This can be solved by the use of an alanine pellet formulation, which closely matches the electron density and mass density of the phantom material. The outcome of the discussions on the energy response of alanine and on the effect of the surrounding materials (and air gaps) on depth dose measurements using alanine pellets in solid phantoms warrant further investigation in order to obtain decisive information on these topics.

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ADDENDUM C: THOMAS ET AL (2003)

THOMAS, R., PALMANS, H., KACPEREK, A. and DUANE, S. Low energy proton beam dosimetry with plane-parallel chambers using NPL electron and ^{60}Co calibrations. *Proceedings of the Workshop on Recent Advances in Absorbed Dose Standards*, ARPANSA, Melbourne, Australia, 19-21 August, 2003.

C.1. Abstract

IAEA TRS 398 [1] recommends that proton dosimetry using ionisation chambers be based on absorbed dose calibrations. However, for protons, as no primary standard exists against which to calibrate chambers, TRS 398 provides factors to derive chamber calibrations in absorbed dose for use in protons from photon beams. Conversion factors from electron to proton beams can be derived as well as shown in this paper. This work aimed to compare measured absorbed dose in the proton beam derived from electron and photon calibrations of the same chamber.

Measurements were conducted in the 60 MeV clinical proton beam at the Clatterbridge Centre for Oncology. Ionisation chambers used were the cylindrical NE2561/2611 and the Scanditronix NACP-02, PTW Roos and PTW Markus parallel plate chambers. All had been calibrated at NPL in terms of absorbed dose, the parallel plate chambers in ^{60}Co and electron beams whilst the cylindrical chambers were calibrated only in ^{60}Co .

Measurements were made in modulated and unmodulated beams. Starting with the same proton beam measurement the absorbed dose for each chamber was derived from the photon and/or electron beam calibration using TRS 398. A transmission monitor was used to limit the influence of beam variations.

In a modulated beam the dose ratio, parallel/thimble, varied between 0.995 and 1.018 dependent on the parallel plate chamber type and choice of starting calibration. For the same beam and starting calibrations, the dose ratio parallel/NACP obtained using Markus and Roos chambers varied between 0.987 and 1.008. In an unmodulated beam (only parallel plate chambers were used) the ratio parallel/NACP varied from 0.986 to 1.019 for Markus and Roos chambers with photon beam and electron beam calibrations.

We conclude that the agreement between the various ionisation chamber types is in general within about 2%, that there is no obviously better consistency with one of both calibration routes and that there is a discrepancy of about 1.5% between both routes which needs further investigation.

C.2. Introduction

Various codes of practice for ionization chamber dosimetry in clinical proton beams have been published in the last decennia; AAPM report 16 [2] and the ECHED code of practice [3, 4] recommend procedures based on air kerma calibrations in a ^{60}Co beam. ICRU report 59 [5] gives procedures for both dosimetry based on air kerma calibrations and absorbed dose calibrations in ^{60}Co , whilst IAEA TRS-398 only contains procedures based on absorbed dose to water calibration factors in ^{60}Co . Though both absorbed dose based codes use a very similar formalism, a number of substantial differences were discovered and mainly attributed to inaccuracies in ICRU report 59 (Medin et al [6]). All the codes mentioned provide data for cylindrical ionization chambers, but, although plane-parallel

chambers are a necessity in the small spread-out Bragg peaks (SOBP) that can occur in low-energy proton beams, only IAEA TRS-398 provides data for these chamber types.

A few publications have compared dosimetry using plane-parallel chambers calibrated in ^{60}Co with dosimetry based on a cross calibration against a cylindrical chamber in a high-energy clinical electron beam or based on a cross calibration against a cylindrical chamber in a proton beam. Wu et al. [7] found all three methods to be in good agreement in a range of proton beam energies (20-170 MeV) for a Markus type chamber using air kerma (or exposure) based procedures, and a PTW-N23333 as reference cylindrical chamber. Another way of expressing this is that both using a ^{60}Co calibration factor and using a cross calibration factor obtained in a high-energy electron beam, the dosimetry in a proton beam with a plane-parallel chamber is consistent with the result from a cylindrical chamber. Medin et al. [8] reported a similar result for an NACP02 type chamber using an NE 2571 as reference. They also used a (prototype) Roos chamber with an absorbed dose to water calibration factor and using an experimentally determined p_{wall} -value of 1.003 in ^{60}Co for this chamber type, they found an agreement within 0.1%. Vatnitsky [9] showed a similar comparison between a Markus-type plane-parallel chamber and an Exradin-T1 cylindrical chamber in a 100 MeV proton beam, using data from ICRU report 59 and air kerma calibrations. He found that based on a cross calibration in a high-energy electron beam, the agreement was within 1.3%, whereas the agreement was better (within 0.4%) when a cross calibration in a high-energy proton beam was used.

Some other publications report comparisons that are based on ^{60}Co calibrations only for the plane-parallel chambers, in general resulting in considerable differences with cylindrical chambers, which are however within the relatively large uncertainties of this method. Palmans et al. [10] compared air kerma and absorbed dose based dosimetry using Markus, NACP02, Calcam-2 and Roos-type plane-parallel chambers and found that both routes gave a similar consistency with cylindrical NE 2571 type chambers (within 1.5%). Kacpersek et al. [11] reported a similar agreement between a Markus and three cylindrical chamber types (Exradin-T1, PTW-30001 and FWT-IC18) based on air kerma calibrations as did Vatnitsky et al. [12] for a Markus chamber versus Exradin-T1, PTW-30001 and NE 2571 chambers based on absorbed dose to water calibrations. A comparison of plane-parallel ionization chamber dosimetry with an independent method was reported by Besserer et al. [13] who performed dosimetry using a Markus type chamber with an absorbed dose to water calibration in ^{60}Co and Fricke dosimetry in 15-25 MeV protons to obtain an agreement within 2%.

Theoretical work on wall perturbation effects due to secondary electron disequilibrium for plane-parallel geometries is also limited to a few references. Laulainen and Bichsel [14] showed analytically that energy transfers by secondary electrons between an aluminium wall and an air cavity are not in equilibrium. This would result in a net energy loss yielding a perturbation correction factor greater than unity. Casnati et al. [15] also used an analytical model to find perturbation correction factors less than unity in a plane-parallel chamber geometry. For example, for a 1mm thick air cavity with A150 walls the (in absolute terms) largest correction would be -2% for proton beams with energies in the range from 1 to 200 MeV. Monte Carlo calculations show, however, that the corrections are more likely to be in the opposite direction [16, 17] although these results are obtained for cylindrical ionization chamber geometries.

NPL offers absorbed dose to water calibration services for both photon [18] and electron [19] beams. The photon service offering calibration for a range of TPR from 0.568 to 0.790 including ^{60}Co and the electron service in the range of $R_{50,D}$ from 1.23 to 6.60 cm. The purpose of the present work is to compare both routes for plane-parallel chamber dosimetry

in a low-energy clinical proton beam. The difference with the work described above is that the electron calibrations are obtained directly from a primary standards laboratory instead of from a clinical beam.

C.3. Materials and methods

C.3.1. Proton beam

Measurements were conducted in the low energy proton beam at the Douglas Cyclotron centre that is part of the Clatterbridge Centre for Oncology, Wirral, UK. The facility was initially commissioned by the Medical Research Council to perform clinical trials of neutron therapy. A proton therapy room was added in 1989 and the facility now treats patients with ocular tumours from all over the world. The beam exits the main facility and is directed in to the treatment room, a diagram of the beam line at this point appears in figure C.1. The proton beam has an energy of 60 MeV with a range in water of 32 mm. The beam profile can then be modified with a collimator on the nozzle, which is visible in figure C.2, to produce various beam shapes up to a maximum diameter of 42 mm. The depth to which the beam penetrates the medium and its depth profile can be modified with range shifters and modulators placed in the beam. The patient chair can be fitted with a platform, as shown in figure C.1, to allow measurement instrumentation to be placed in the beam. For this experiment we used a beam diameter of 30 mm and dose rates of about 9 Gy min^{-1} for the modulated beam and 13 Gy min^{-1} for the unmodulated beam.

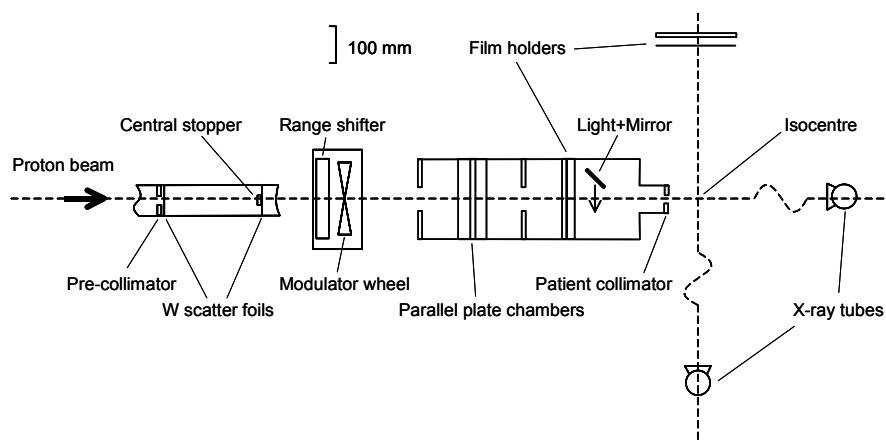


Figure C.1. A simplified diagram of the proton beam line on entry in to the patient treatment room of the Douglas Cyclotron (CCO) [Adapted from ref 31].

C.3.2. Ionization chambers, phantom and calibration modalities

Dosimetry using six plane-parallel chambers (two NACP-02, two Markus and two Roos chambers) was performed in both the modulated and unmodulated 60 MeV proton beam. In the modulated beam the NPL designed NE 2561 and the NE 2611 cylindrical chamber, which are recommended for use as the photon therapy level secondary standard chambers in the UK were used as a reference. Whilst it is widely accepted that these two versions of the same basic chamber type have no differences it was felt useful to examine each version in the proton beam.

These chambers were placed in a 26 cm cubic water filled phantom as shown in figure C.2. The phantom is in fact the one that is used for the NPL electron absorbed dose calibration service. The phantom is constructed of 1 cm thick Perspex walls with a 2 mm thick, 15 cm diameter front window. Slots positioned in the phantom allow the chambers to be positioned at 7 basic depths designed to match those required to correctly position the chambers for the reference depths of the NPL electron beams. Use of additional spacers allows the positioning of the front window of the chamber to be varied over a wider range of depths from approximately 3mm.

The chambers were, as far as possible, calibrated using the standard procedures used for the photon and electron calibration services. The Primary standards for both the photon and electron calibration services are graphite calorimeters. The photon primary standard graphite micro calorimeter is based on the design by Domen and Lamperti [20] and is described in detail in a number of publications [21]. The electron calorimeter is of a much simpler construction and is described by McEwen et al [22]. In both cases a set of transfer chambers are calibrated against the calorimeters firstly in terms of absorbed dose to graphite and then using the photon fluence scaling theorem, as described by Nutbrown et al [23] and McEwen et al [24], converted to absorbed dose to water. These transfer chambers are then used to calibrate the secondary standard chambers in various beams by the method of direct replacement.

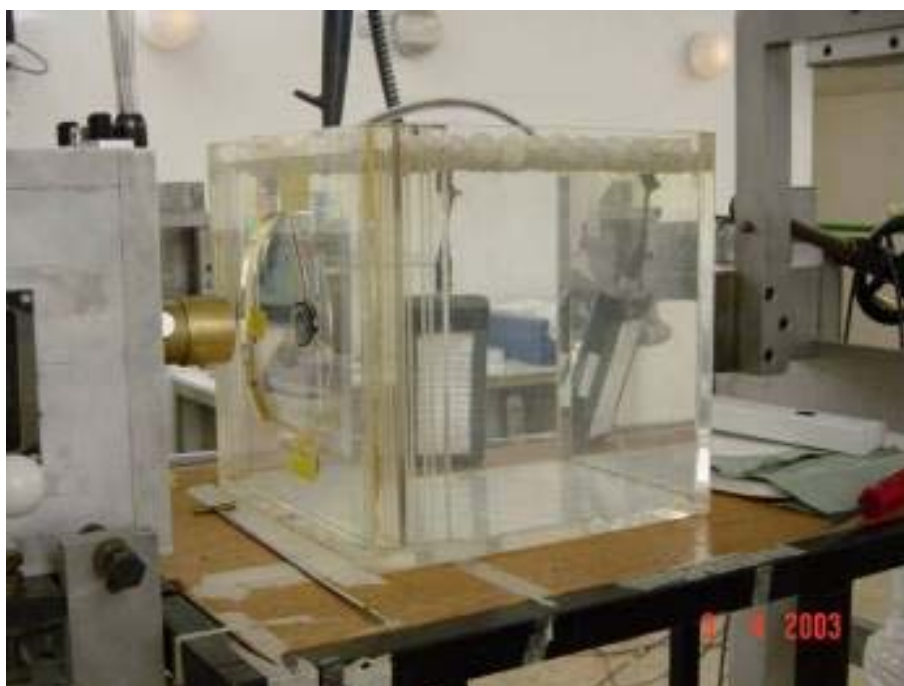


Figure C.2. The water filled phantom set up in the treatment room of the proton facility at Clatterbridge. A Markus chamber can be seen in the phantom just right of the brass nozzle and collimator (the water is covered with a layer of balls to minimize evaporation and assist temperature stability).

Departures from the standard calibration procedures at NPL arose due to the following reasons:

1. The NE 2561/2611 chamber is not recommended for use in electron beams and therefore was not calibrated in that modality.

2. The Markus chamber is not recommended for reference electron dosimetry in the UK, however the calibration of the chamber was carried out in the same way.
3. Although at NPL plane parallel electron chambers are not routinely calibrated in ^{60}Co photon beams this has previously been documented by McEwen et al [25] and a similar procedure was carried out here.

C.3.3. Determination of absorbed dose to water for a proton beam using TRS 398

The IAEA absorbed dose code of practice TRS 398 allows for the conversion of absorbed dose calibration factors made in both electron and photon beams to be converted for use in proton beams.

Starting with a calibration factor N_{D,w,Q_0} in a reference beam quality Q_0 the measured absorbed dose is found from:

$$D_{w,Q} = M_Q N_{D,w,Q_0} k_{Q,Q_0} \quad (\text{C.1})$$

where k_{Q,Q_0} is the theoretically derived factor, which converts the calibration factor from Q_0 to Q and which is for the particular chamber type given in Table 31 of TRS 398. If Q_0 is ^{60}Co then this index is omitted.

For chambers with an electron absorbed dose calibration factor the conversion to absorbed dose for protons is a two-step procedure. Firstly the electron factor is converted to a ^{60}Co factor using Table 18 of TRS-398 and then to a proton beam factor via table 31. Assuming the W_{air} value for ^{60}Co is the same as for the 19 MeV electron calibration beam quality Q_{e19} , $k_{Q,Q_{e19}}$ is given by:

$$\begin{aligned} k_{Q,Q_{e19}} &= \frac{(W_{air})_Q (s_{w,air})_Q P_Q}{(W_{air})_{Q_{e19}} (s_{w,air})_{Q_{e19}} P_{Q_{e19}}} \\ &= \frac{(W_{air})_Q (s_{w,air})_Q P_Q}{(W_{air})_{^{60}\text{Co}} (s_{w,air})_{^{60}\text{Co}} P_{^{60}\text{Co}}} \cdot \frac{(s_{w,air})_{^{60}\text{Co}} P_{^{60}\text{Co}}}{(s_{w,air})_{Q_{e19}} P_{Q_{e19}}} \\ &= k_Q \cdot \frac{1}{k_{Q_{e19}}} \end{aligned} \quad (\text{C.2})$$

Thus, ^{60}Co acts here as an intermediate beam quality such as is used for electron beams in TRS 398 (which has no relation to any physical measurement) and is basically only introduced for making the tabulations of data in TRS-398 more concise..

With chambers that have calibrations for both ^{60}Co and electron beams this allows the absorbed dose in the proton beam to be calculated via these two routes using the same measurement made in the proton beam.

The factors that were used in the calculation of absorbed dose to water for the proton beam are shown in Table C.1.

Chamber Type	Chamber serial number	$N_{D,w,Co}$	$N_{D,w,e}$	k_Q proton beam R_{res} 1.5 g/cm ² (table 31)	k_Q electron beam R_{50} 6.6 g/cm ² (table 18)	$N_{D,w,p(Co)}$	$N_{D,w,p(e)}$	$N_{D,w,p(Co)}$
								$N_{D,w,p(e)}$
NE2561	289	0.1029	-	1.0380	-	0.1068	-	-
NE2561	189	0.1026	-	1.0380	-	0.1065	-	-
NACP02	5205	0.1652	0.1493	0.9890	0.8884	0.1634	0.1662	0.983
NACP02	6306	0.1544	0.1397	0.9890	0.8884	0.1527	0.1555	0.982
Roos	227	0.0831	0.0751	1.0020	0.9068	0.0832	0.0829	1.004
Roos	133	0.0826	0.0746	1.0020	0.9068	0.0828	0.0824	1.005
Markus	478	0.5568	0.5048	1.0030	0.8960	0.5585	0.5651	0.988
Markus	2225	0.5329	0.4831	1.0030	0.8960	0.5345	0.5408	0.988

Table C.1. Factors used for the conversion of calibrations in Cobalt 60 ($N_{D,w,Co}$) and 19 MeV electrons ($N_{D,w,e}$) for the calculation of absorbed dose to water in the proton beam using TRS-398.

It would be expected for the ratio of the ⁶⁰Co to 19 MeV factors for individual parallel plate chambers of the same type to be similar, as can be seen in the final column of Table C.1.

From the three last columns it is also obvious that both routes do not yield the same absorbed dose calibration factor for protons, indicating that there is an inconsistency between the data in TRS-398 and the ratio of the calibration factors.

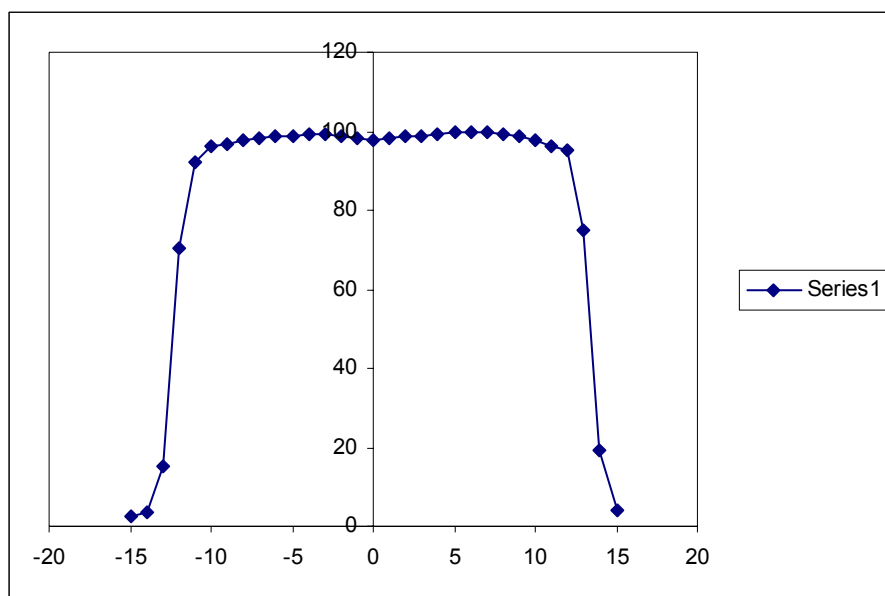


Figure C.3. Lateral plot of beam uniformity measured using CCO diode.

The plot in figure C.3 shows an example of the beam profile. It can be seen that there is a dip in the central region of the beam of the order of 1 to 2 %, since the diameters of the chambers varied considerably it was necessary to correct for this variation.

The results given here are not corrected for ion recombination or polarity. Some preliminary measurements to calculate the ion recombination correction for the NE 2561/2611 and the NACP02 chambers were made, these provisional measurements suggest the recombination correction will not be greater than 0.3 %.

Figure C.4 shows percentage depth dose distributions for the modulated and unmodulated beams measured using a diode. The arrows on the figure show the approximate reference depths at which the ion chamber measurements were performed.

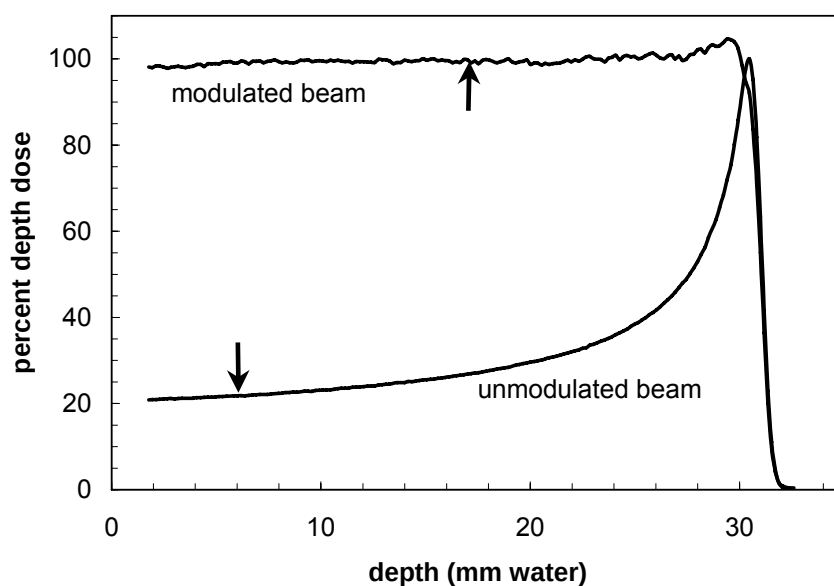


Figure C.4. Percent depth dose distributions for the modulated and unmodulated beams.

C.4. Results

Figures C.5 and C.6 show results for the modulated beam, 7 and 8 the unmodulated beam. Each point represents the ratio of the measurement made with the individual chamber to that of the mean of the absorbed dose measured by either the NE 2561/2611 cylindrical chambers (cyl) or the NACP02 parallel plate chambers (pp).

The cylindrical chambers were not used in the unmodulated beam due to their relatively large physical size when compared to the large dose gradient of the proton beam.

Figure C.5 shows ratios for each chamber against both the cylindrical and parallel plate reference chambers using the ^{60}Co derived calibration. Results show agreement within 2% with the Markus chambers being below and the Roos chambers being above the average in relation to both sets of reference chambers. There is good agreement between the two sets of reference chambers, even though the individual NACP chambers show a considerable discrepancy of the order of 1.5%.

Figure C.6 shows ratios for each chamber against both the cylindrical and parallel plate reference chambers using the 19 MeV derived calibration, however the ^{60}Co calibration was

used for the reference cylindrical chambers. Within each “set” there is agreement at the level of 1% with one outlier taking this to 2.5%. There is a difference of approximately 1.6% between the reference chambers, suggesting a discrepancy between the two routes (the ^{60}Co photon and 19 MeV electron calibration) used to derive the calibration factors. Now both the Roos and Markus chambers are below the NACP02 chambers.

Figure C.7 shows, for the unmodulated beam, ratios for each chamber against the parallel plate reference chambers using the ^{60}Co derived calibration. The maximum difference between individual chamber results is of the order of 2.5 % with the Roos chambers appearing high by 1.4% and the Markus chambers low by 0.7% when compared with the mean of the reference chambers.

Figure 5 Chamber ratios for modulated beam based on ^{60}Co photon calibration

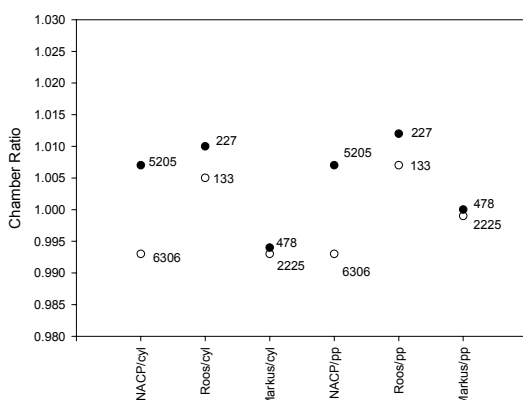


Figure 6 Chamber ratios for modulated beam based on 19 MeV electron calibration (reference cyl chambers using ^{60}Co factor)

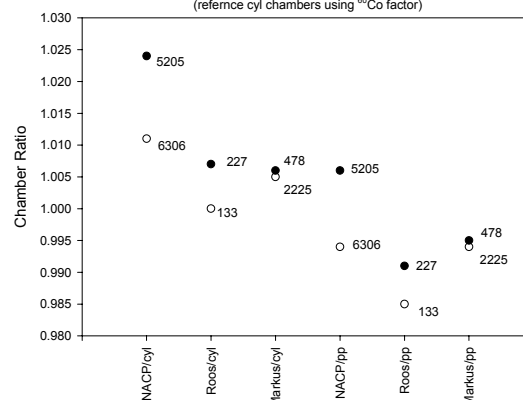


Figure 7 Chamber ratios for unmodulated beam based on ^{60}Co photon calibration

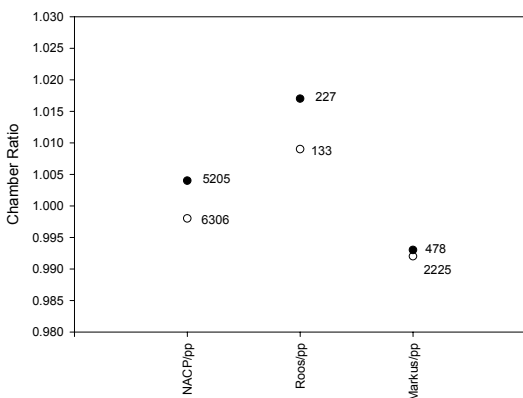
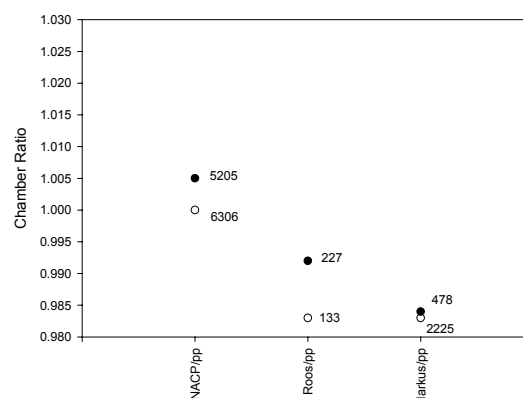


Figure 8 Chamber ratios for unmodulated beam based on 19 MeV electron calibration



Figures C.5. to C.8.

Figure C.8 shows ratios for each chamber against the parallel plate reference chambers using the 19 MeV derived calibration. The maximum difference is now of the order of 2% with the Markus chambers remaining low by about 1.5% and the Roos chambers below the reference to agree within 1.2% of the reference chambers.

Table C.2 shows estimates for the combined standard uncertainties on the absorbed dose measurements with any of the plane-parallel ionisation chambers used in this work. For cylindrical chambers, the uncertainty on k_Q could be slightly smaller [1]. The last two columns show uncertainty estimates on the ratio of k_Q -values that could be derived from the ratio of two chamber readings. The uncertainty on the ratio of calibration factors that is involved here is reduced because a number of factors specific to the primary standards

cancel out. The latter uncertainties can be used to evaluate if the ratio of theoretical k_Q values are inconsistent with experimental values within certain confidence limits.

	u_c based on ^{60}Co (%)			
	D_w		k_Q -ratio	
	Non-modul.	Modul.	Non-modul.	Modul.
calibration (A+B)	0.70	0.70	0.24	0.24
charge+pT (A)	0.1	0.1	0.14	0.14
reprod (A)	0.2	0.2	0.28	0.28
pion (A+B)	0.35	0.35	0.17	0.17
positioning (B)	0.22	0.07	0.31	0.10
lateral profile (B)	0.3	0.3	0.3	0.3
k_Q (B)	2.0	2.0		
Combined	2.2	2.2	0.6	0.5

	u_c based on 19MeV electrons (%)			
	D_w		k_Q -ratio	
	Non-modul.	Modul.	Non-modul.	Modul.
calibration (A+B)	0.75	0.75	0.42	0.42
charge+pT (A)	0.1	0.1	0.14	0.14
reprod (A)	0.2	0.2	0.28	0.28
pion (A+B)	0.35	0.35	0.17	0.17
positioning (B)	0.22	0.07	0.31	0.10
lateral profile (B)	0.3	0.3	0.3	0.3
k_Q (B)	1.6	1.6		
Combined	1.8	1.8	0.7	0.6

Table C.2. Estimated uncertainties for the measurement of absorbed dose to water D_w with an ionisation chamber in this work and on the ratio of k_Q values for two ionisation chambers derived from a ratio of chamber readings.

C.5. Discussion and Conclusions

For the ^{60}Co derived factors (figure C.5) there is generally good agreement with maximum mutual difference between all the chambers of the order of 1.8%. The fact that the discrepancy of about 1.5% between the two NACP chambers is the same using the ^{60}Co and the electron calibrations indicates that it is directly related to the proton measurements. It is worth noting that we found the response of the NACP chambers to be very sensitive to the lateral positioning, with a 2mm lateral displacement resulting in a 3% change in response. This is possibly a result of the dip in the lateral profiles, shown in figure C.3 and contributes to the overall uncertainty. The results in figure C.6 show that there is a difference in the two calibration routes of the order of 1.6%. The primary standard calorimeters on which each

service is based have been compared and have been shown to agree within 1%, but both procedures require a conversion procedure from graphite to water that may not be entirely consistent. Another source of this inconsistency could arise from the data in TRS-398. This issue was raised before [31] and is currently under investigation at NPL.

Measurements made in the unmodulated beam shown in figure C.7 and figure C.8 show the largest difference between the two routes for the Roos chamber of 2.4%. Because of the large change in dose gradient in the unmodulated proton beam it could be suggested that this is due to a positioning error on the phantom, however, the same measurements made in the modulated beam show a difference of 1.5%. This could be partially related to the value of 1.010 assigned to p_{wall} in ^{60}Co for the Roos chamber in TRS-398, investigations indicate that on average the experimental values are about 1% higher [27,28,29 and 30] which would bring the ^{60}Co results in Figures C.5 and C.6 down.

Overall, the results show that both calibration routes show the same level of consistency, but that differences for individual chambers could be up to 1.7% as shown in Table C.1. Further work is necessary to measure and correct for the effects of ion recombination and to investigate the source of discrepancies between the NACP chambers.

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ADDENDUM D: PALMANS AND VERHAEGEN (2005)

PALMANS, H. and VERHAEGEN, F. Assigning nonelastic nuclear interaction cross sections to Hounsfield Units for Monte Carlo treatment planning of proton beams. *Phys. Med. Biol.*, 2005, **50**, 991-1000.

D.1. Abstract

In high energy clinical proton beams nonelastic nuclear interactions contribute substantially to the total dose. It is therefore of importance to know these contributions quantitatively and to be able to scale them correctly as a function of Hounsfield Units obtained from CT data.

In this work, the second of these issues has been addressed. The importance of taking material dependent nonelastic nuclear interactions into account has been investigated for Monte Carlo calculations. A scaling curve for nonelastic nuclear interactions as a function of Hounsfield Unit has been established and compared with similar data for the stopping powers. Monte Carlo simulations using McPTRAN.MEDIA and MCNPX have been performed in homogenous media and in inhomogeneous slab geometries.

The results show that for skeletal tissues and for adipose tissue, the tissue to water nonelastic cross section ratios differ up to 10% compared to the tissue to water stopping power ratios. This results in errors of the order of 2-3% when both contributions to the total dose are scaled in the same way (with stopping power ratios). Monte Carlo simulations in slab geometries with tissue materials for 200 MeV protons show similar effects, but when both contributions are scaled correctly the errors are no larger than 0.5% in the situations investigated here.

D.2. Introduction

Due to the ballistic properties of protons, and the transport of secondary particles along the beam axis inherently taken into account by using measured depth dose curves, pencil beam models based on Fermi-Eyges theory usually give satisfactory results in proton dose calculations. Nevertheless, especially in the presence of lateral inhomogeneities, Monte Carlo simulations play an important role in treatment planning as well as in studying the water equivalence of various tissues. Of concern in Monte Carlo treatment planning (MCTP) for protons is the quantitative contribution of nonelastic nuclear interactions to absorbed dose. At high energies, these interactions contribute substantially while their cross sections are poorly known. ICRU report 63 (ICRU 2000) provides until now the most comprehensive compilation of data, but quotes standard uncertainties of 5 to 10 % on the total nonelastic cross sections for protons.

Some authors have quantified the effect of secondary charged particles resulting from nonelastic nuclear interactions on stopping powers showing that these can contribute up to 20% at the highest energies used clinically (Laitano et al. 1996, Paganetti 2002). It has also been shown that the transport of secondary protons gives rise to a substantial build-up effect at the surface (Laitano et al. 1996, Medin and Andreo 1997, Paganetti 2002). The effect of including transport of secondary protons on stopping power ratios, on the other hand, is negligible as shown qualitatively by Laitano et al. (1996) and quantitatively by Medin and Andreo (1997) and Laitano and Rosetti (2000).

In MCTP in general, stopping powers or electron densities are assigned to Hounsfield Units (HU) obtained from Computed Tomography (CT) by assuming a scaling model (see e.g.

Kawrakow et al. 1996 for electron dose calculations) or by developing a model for deriving the material compositions from the HU (du Plessis et al. 1998). For proton beams, the relation between stopping powers and HU has been investigated by Schneider et al. (1996) and Schaffner and Pedroni (1998) and for heavy ions a similar approach has been used by Jäkel et al. (2001).

One of the issues is that x-ray interaction probabilities and proton interaction probabilities are not the same. Methods that have been tried to overcome this are dual CT (Torikoshi et al. 2003 for heavy ions) and proton CT (Schneider and Pedroni, 1995). For the stopping powers it is in general not problematic since they vary from material to material in a similar way as for electrons, but for nonelastic nuclear interactions this is a potential issue. Matsufuji et al. (1998) and Torikoshi et al. (2003) have addressed this problem for heavy ion beams where it is definitely a more important issue due to the significant role nuclear interactions play.

In this work, the importance of taking material dependent nonelastic nuclear interactions into account for protons is investigated. A scaling curve for nonelastic nuclear interactions as a function of HU is established in a similar way as was done by Schneider et al. (1996) for the stopping powers. Monte Carlo calculations using McPTRAN.MEDIA and MCNPX are performed in homogenous media and in inhomogeneous slab geometries. These are based on ICRU specified tissue compositions.

D.3. Nonelastic interactions compared to electromagnetic interactions

D.3.1. Importance as a function of energy

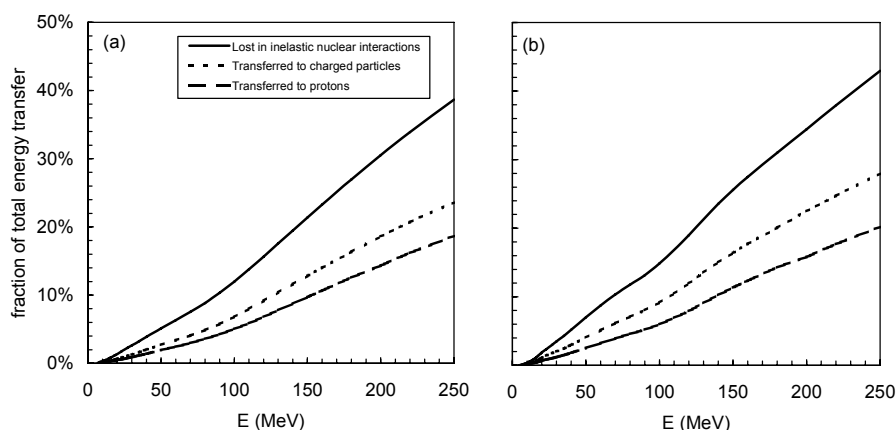


Figure D.1. Fractions of the total energy transferred by protons in nonelastic interactions as a function of energy (a) in water and (b) in ^{12}C .

The stopping powers are taken from ICRU 49, the nonelastic nuclear interaction cross sections from ICRU 63. Figure D.1 shows for both water and carbon the fractions of the total energy transfer:

- (i) that is lost in nonelastic nuclear interactions
- (ii) that is transferred to charged ions (including protons)
- (iii) that is transferred to protons

This shows that for energies below 60 MeV (the energy range used for ocular tumours) the nonelastic interactions constitute only a few percent of the total energy transfer and thus it can be expected that for these low-energy beams the effects related to nonelastic interactions will be small. In the high-energy range (above 150 MeV), on the other hand, the contributions of nonelastic interactions to the total energy transfer are considerable. In water for example, the contribution that is transferred to charged heavy particles is around 20% for a 200 MeV proton.

D.3.2. Discussion for tissues and tissue substitutes as a function of Hounsfield Unit

Nonelastic nuclear interaction cross sections were calculated for the same list of materials as used by Schneider et al. (1996) for establishing a calibration curve for the stopping powers as a function of scaled HU. These were calculated as described by Schneider et al. (1996) and normalised to 1000 for water. The materials were taken from ICRU report 44 (1989) and ICRP report 23 (1975) and consist of tissues and tissue substitutes. The total nuclear interaction cross section per atomic mass unit (or the production cross sections for a particular secondary particle per atomic mass unit) in a compound is calculated from the elemental cross sections as

$$\left(\frac{\sigma}{A}\right)_{\text{compound}} = \sum_i w_i \cdot \left(\frac{\sigma}{A}\right)_i \quad (\text{D.1})$$

where w_i are the fractions by weight of the i^{th} element.

ICRU 63 provides data for the most important elements in tissues: carbon, nitrogen, oxygen, phosphorus and calcium. The abundance of other elements (sodium, magnesium, sulphur, chlorine, potassium, iron and iodine) in the tissues that are considered here is always smaller than 1% and mostly smaller than 0.3% by weight. Since for these elements no data are given in ICRU 63 they were obtained by interpolating the cross sections (both the total and partial cross sections) as a function of the logarithm of the atomic number. The errors induced by this approximation can be considerable for a particular element but are supposed to be negligible given the small concentrations of these elements.

For tissue substitutes, however, the situation is different since for some formulations the concentration of elements like magnesium, sulphur, chlorine or fluorine is more substantial. Hence, care should be taken with interpreting the results based on the tissue substitute data as well as on the choice of tissue substitutes when they would be used for calibrating CT HU for proton MCTP calculations.

Figure D.2 shows the relative nonelastic cross section per unit of volume for 219 MeV protons as a function of scaled HU, with water as reference. This energy was the same as used by Schneider et al. (1996) for their analysis and experiments and is conveniently high since it is in the energy range above 200 MeV that effects are expected to be large. There is a considerable energy dependence but this is especially the case at low energies where the contribution of nonelastic nuclear interactions to the total dose is small. A curve representing the relative stopping powers from Schneider et al. (1996) is shown as well. These data were based on the mean excitation energies from Janni (1982). We have verified that recalculating the relative stopping powers using mean excitation energies from ICRU 49 does not result in significant differences. It is immediately clear that in the region of HU units between 1250 to 2500, the curve representing the nonelastic cross sections for all

skeletal tissues is different from the relative stopping power curve. This will result in an error when the contributions to absorbed dose from nonelastic cross sections are scaled with the relative stopping powers. Also obvious is that for non-tissue materials, the deviations can be very large, an effect that also shows up, although to a lesser extent, for the relative stopping powers (Schneider et al. 1996). This shows that care should be taken in selecting tissue substitutes for calibrating HU units for proton treatment planning. Teflon for example would be a bad idea, whereas RMI lung and RMI bone are in good agreement with the tissue curve. From the detailed graph it appears that for adipose tissue and yellow bone marrow, the relative nonelastic cross sections also deviate considerably from the relative stopping power. For these reasons, we can expect the largest effects for adipose tissue and cortical bone and these tissue materials have been selected for a more detailed Monte Carlo simulation below.

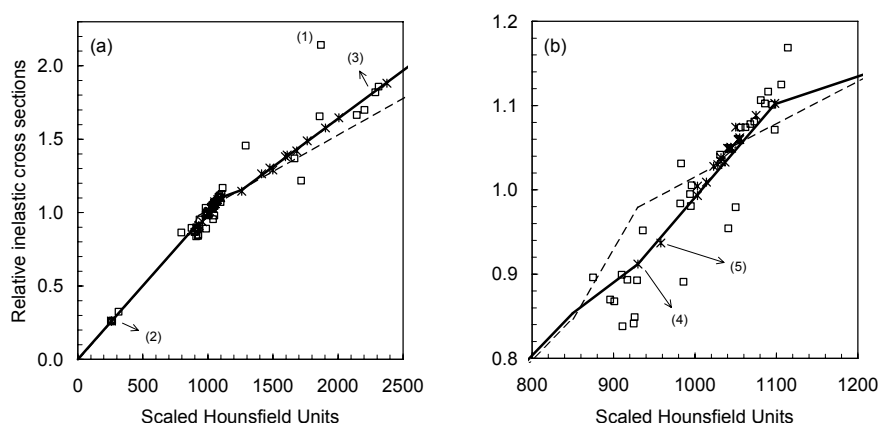


Figure D.2. Relative nonelastic cross section per unit of volume for 219 MeV protons as a function of scaled Hounsfield Unit, with water as reference using ICRU 63 data (a) for the whole HU range and (b) detail for the HU range between 800 and 1200. The cross symbols are for tissues, the square symbols for tissue substitutes. The solid line is a calibration curve based on the tissues only. For comparison, the calibration curve for the relative stopping powers from Schneider et al. (1996) is represented as a dashed curve. The labels refer to Teflon (1), RMI lung (2), RMI bone (3), adipose tissue (4) and yellow bone marrow (5).

Similar plots for the fraction of the energy in nonelastic nuclear interactions that is transferred to charged particles lead to the same observations. Another way of looking at the differences between nonelastic interaction cross sections and stopping powers is by plotting them against each other as is done in figure D.3a. Again, the deviation for skeletal tissues is obvious. Figure D.3b, presenting similar information based on the data set of Janni (1982), reveals similar qualitative effects as for the ICRU 63 data but shows a strong quantitative dependence on the data set.

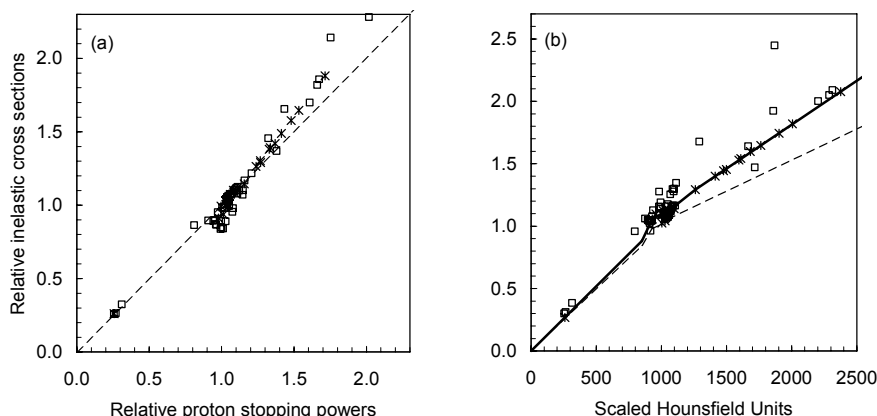


Figure D.3. (a) Relative nonelastic cross section per unit of volume for 219 MeV protons plotted as a function of relative stopping power per unit of volume and (b) relative nonelastic cross section per unit of volume for 219 MeV protons as a function of scaled Hounsfield Unit, with water as reference using Janni (1982) data.

D.4. Monte Carlo simulations

D.4.1. Methods

Simulations were performed with two Monte Carlo codes: MCNPX (Waters 2002) and McPTRAN.MEDIA (Palmans 2004), which is an in-house, modified version of PTRAN (Berger 1993). The latter only models the transport of mono-energetic, monodirectional pencil beams in a homogeneous water geometry, whereas McPTRAN.MEDIA extends it with the potential of simulating proton transport in other materials than water in non-homogeneous slab geometries, for beams with lateral, angular and energy distributions as well as modulated proton beams. These various features have been described in several publications (e. g. Palmans and Verhaegen 1997, 1998) and have been summarised by Palmans (2004). For this work, McPTRAN.MEDIA and MCNPX simulations have been performed in homogeneous and non-homogeneous slab geometries for mono-energetic proton pencil beams incident perpendicularly to the slabs. Apart from potential differences in the interaction data used by both codes, the major difference is that in MCNPX secondary particles from nonelastic nuclear interactions are further transported whereas in McPTRAN.MEDIA they deposit their energy at the location where they are generated.

Calculations of depth dose distributions in a homogeneous geometry were performed for adipose tissue, skeletal muscle, striated muscle, compact bone, cortical bone and water according to the elementary compositions given in ICRU report 49. These are the same as used by Schneider et al (1996) and in the previous discussion. The calculations were done for 60 MeV and 250 MeV mono-energetic proton beams. Subsequently, simulations for a 50 mm bone slab at a depth of 50 mm in water and a 50 mm adipose slab in water at a depth of 2.5 mm were performed for 200 MeV protons.

D.4.2. Results in homogeneous materials

Figures D.4a and D.4b show the ratio of the calculated absorbed dose in the various tissue compositions and the absorbed dose at an equivalent depth, multiplied with the stopping power ratio water to tissue, $S_{w,tissue}$:

$$(D/S_C)_{tissue,w} = D_{tissue} \cdot S_{w,tissue} / D_w \quad (D.2)$$

Equivalent depths scale with the ranges r_0 in the continuous slowing down approximation (csda). The factors shown in figure D.4 are the reciprocals of the fluence correction factors as defined by Palmans et al. (2002) although it could be argued that since the difference of the proton fluence in water and tissue at equivalent depths is mainly the result of a difference in nonelastic nuclear attenuation it is predominantly an attenuation correction factor as defined by Laitano et al. (1996). If the depth dose functions in tissue material and water would be simply scalable with the csda ranges and the stopping power ratios, then this ratio should be unity.

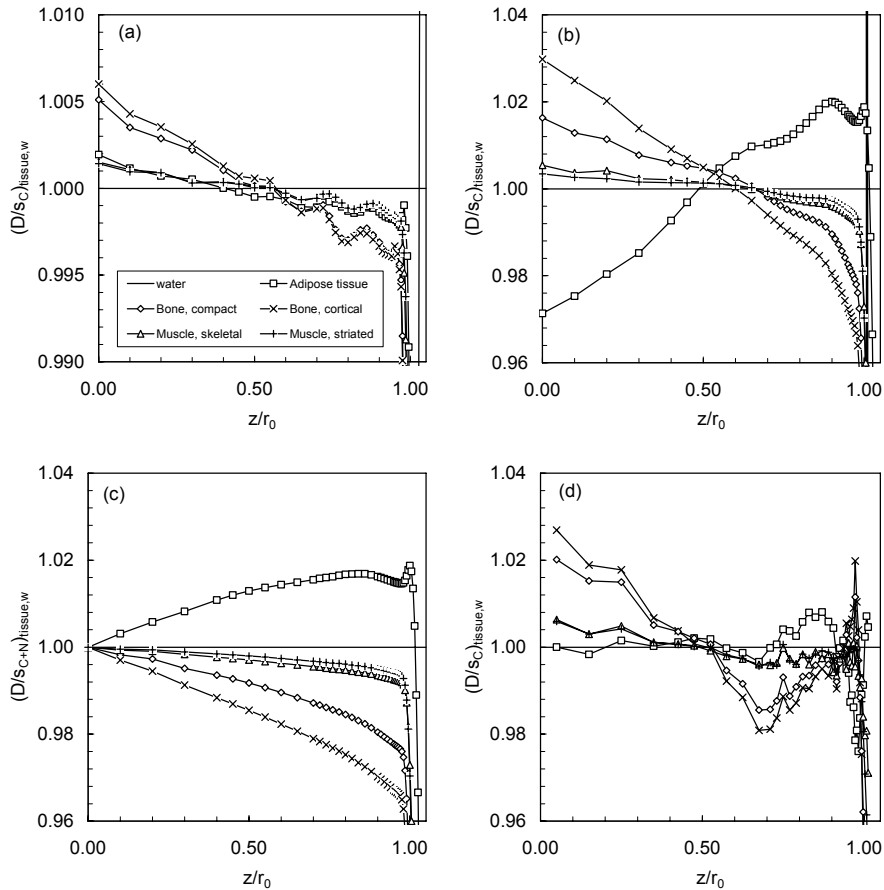


Figure D.4. The quantity of expression (2) for 60 MeV protons (a) and for 250 MeV protons (b) and the quantity of expression (3) for 250 MeV protons (c) calculated using McPTRAN.MEDIA. Figure (d) shows similar results as figure (b) calculated with MCNPX for 250 MeV protons.

One issue illustrated in figures D.4a and D.4b is that the contributions to the dose in nonelastic nuclear interactions, $D_{\text{tissue},N}$, should be converted separately from the dose deposited in Coulomb interactions, $D_{\text{tissue},C}$. This can be derived from the observation that the ratio is not unity at the surface when only the electromagnetic stopping power ratios are used, as it should be since there is no difference in fluence at the surface (given the negligible amount of backscatter protons undergo). In figure D.4c this is demonstrated for 250 MeV protons where the ratio has been calculated as:

$$(D/s_{C+N})_{\text{tissue},w} = (D_{\text{tissue},C} \cdot s_{w,\text{tissue}} + D_{\text{tissue},N} \cdot (E \cdot \sigma/A)_{w,\text{tissue}}) / D_w \quad (\text{D.3})$$

Note that the factor “ E ” in the second term on the right hand side in equation (D.3) was not explicitly included in a similar expression by Palmans et al. (2002). In the latter publication an equivalent energy was assigned to a given residual range (Andreo et al. 2000) and it was assumed that the equivalent energy at equivalent depths in both phantom materials was the same such that this factor cancelled out in the ratio. The expression given here is more general in the sense that it allows for differences in energy spectra at equivalent depths and it also reflects the difference in the way it has been evaluated, i.e. in-line in the Monte Carlo simulations.

Inspecting figure D.4a we can see that for a low-energy beam it is a reasonable assumption to convert doses based on the stopping powers only, the errors due to an incorrect conversion being limited to 0.5%. Nonelastic nuclear attenuation has only a small effect, which could be anticipated from the discussion of figure D.1 regarding the low-energy range.

Figure D.4b shows that for high-energy beams, more care should be taken with respect to the correct conversion of the two contributions. Especially for cortical bone and adipose tissue, converting both components with the stopping powers results in errors of the order of 3%. This could be expected given the discussion in section D.2.2. Figure D.4c shows that nonelastic nuclear attenuation over an (unrealistically large) range of 35 cm leads to differences of about 2% for the same tissues. These observations indicate that it is relevant to take both contributions into account separately. The MCNPX results plotted in figure D.4d show that for both types of bone and muscle the two codes agree reasonably well except near the end of the range. In this region (the Bragg peak) the results are sensitive to slight deviations in range scaling. Scaling the elastic and nonelastic contributions separately will yield similar conclusions as for McPTRAN.MEDIA although the effects will be slightly smaller and limited to 2%. Note that near the entrance region, no substantial build-up related difference between McPTRAN.MEDIA and MCNPX is visible. This indicates that, although secondary proton transport is important for knowing the absolute dose distribution accurately, it has a negligible influence on scaling dose distributions from one material to another. This allows us to trust the McPTRAN.MEDIA results as much as the MCNPX results in the quantitative study of dose conversion factors from one medium to another, even if in McPTRAN.MEDIA secondary protons are not transported and thus the absolute dose distribution is not accurate as demonstrated by Medin and Andreo (1997). These observations are in agreement with the findings of Medin and Andreo (1997) and Laitano and Rosetti (2000) regarding stopping power ratios, which are explained by the slow variation of stopping power ratios with energy. For adipose tissue, there is a substantial discrepancy between both codes and unlike for the McPTRAN.MEDIA result, the MNCPIX result is close to unity. This means that in MCNPX, the relative nonelastic nuclear interaction cross sections are close to the relative stopping powers for adipose tissue. The

effect is found to be due to the limit of the nonelastic nuclear interaction database in MCNPX, which goes only up to 150 MeV. A McPTRAN.MEDIA simulation in which the nonelastic nuclear interaction cross sections for higher energies are kept to the value at 150 MeV confirms this. It can be understood by noticing that in the ICRU 59 data the cross sections per nucleon for ^{16}O are almost constant above 150 MeV whereas the cross sections per nucleon for ^{12}C , being close to the oxygen values at 150 MeV, still vary considerable above this energy (ICRU 1998) and for adipose the carbon content happens to be much larger than in the other tissues.

D.4.3. Result for a slab of tissue in water

Figure D.5 shows the same quantity as plotted in figure D.4 but for 5 cm water thickness equivalent slabs of cortical bone and adipose tissue in water for a 200 MeV mono-energetic proton beam.

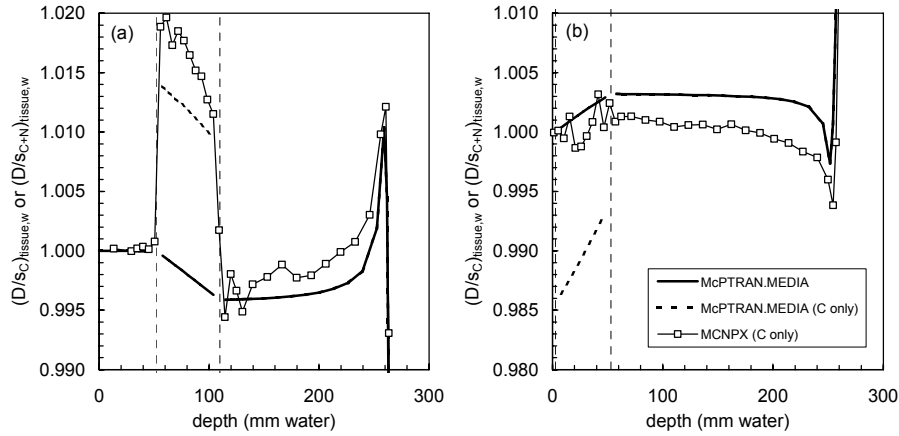


Figure D.5. The quantities of expressions (2) (only for McPTRAN.MEDIA) and (3) (for both McPTRAN.MEDIA and MCNPX) for 5 cm water equivalent slabs of (a) cortical bone and (b) adipose tissue.

The results obtained with McPTRAN.MEDIA show that if a correct conversion is done for both contributions, the deviations from unity are small: less than 0.5% down to almost the end of the range for both situations. When all contributions are converted with expression (2) the differences can again be substantial in both the McPTRAN.MEDIA and MCNPX simulations: 1.5% to 2% for cortical bone and about 1% for adipose tissue. Again, the McPTRAN.MEDIA and MCNPX results do not agree very well for adipose tissue; figure D.5b indicates that both contributions scale in the same way in MCNPX but not in McPTRAN.MEDIA (the latter essentially reflects the ICRU 49 and ICRU 63 data). We can observe that the build up effects near material interfaces have no effect in the ratios shown in figure D.5, supporting the conclusion in the previous section on this issue.

D.5. Conclusions

The importance of taking material dependent nonelastic nuclear interactions to the total dose into account in proton beams has been investigated. A scaling curve for nonelastic nuclear interactions as a function of HU has been established and compared with similar data for the

stopping powers. This shows that a separate scaling curve from the one used for stopping powers is useful, especially for skeletal tissues and adipose tissue for which the difference between the relative nonelastic cross sections and the relative stopping powers is of the level of 10% based on ICRU 63 data. It also shows that care should be taken in the selection of potential tissue substitutes for calibrating a CT curve, since some materials deviate considerably.

Monte Carlo simulations using McPTRAN.MEDIA and MCNPX in homogeneous media show that errors of the level of 2-3% are made when both contributions of nonelastic nuclear and Coulomb interactions to the total dose are scaled in the same way (with stopping power ratios). Monte Carlo simulations using McPTRAN.MEDIA and MCNPX for 5 cm water thickness equivalent slabs of cortical bone and adipose tissue in water show similar effects, but when both contributions are scaled correctly the errors are no larger than 0.5% in the situations investigated here.

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ADDENDUM E: SHIPLEY ET AL (2005)

SHIPLEY, D.R., PALMANS, H., BAKER, C. and KACPEREK, A. Geant4 simulation of an ocular proton beam and benchmark against other Monte Carlo codes. *In: The Monte Carlo Method: Versatility Unbounded in a Dynamic Computing World* (Illinois, USA: American Nuclear Society), 2005.

E.1. Abstract

This paper describes recent work using GEANT4 to simulate monoenergetic proton beams (50 MeV, 150 MeV and 250 MeV) and the 62 MeV ocular proton beam line at the Clatterbridge Centre for Oncology (CCO). GEANT4 results are compared with those obtained using other Monte Carlo codes (MCNPX, PTRAN, and SRIM) and measurement, including depth-dose data and radial dose and energy distributions at two different depths.

GEANT4 and PTRAN are in generally good agreement but differences in both the depth and height of the Bragg peak become more apparent at higher energies due to the parameterisation in GEANT4 of stopping power data taken from ICRU 49, and by some differences in their inelastic nuclear interaction cross-section data. Other variations at the energies under consideration are accounted for by PTRAN's lack of secondary proton transport. GEANT4 is in better agreement with MCNPX at all energies but again differences in the nuclear interaction data result in MCNPX giving a higher relative dose compared to GEANT4 at depths below the Bragg peak. The different multiple scattering model used in MCNPX also tends to give broader radial dose distributions and significantly larger tails in the energy distributions in water compared to GEANT4.

Further simulations of depth dose and radial dose distributions in PMMA are compared with diode and film measurements on the CCO ocular proton beam line. GEANT4 overestimates the height of the Bragg peak for the full-energy beam by around 20% and gives a slightly increasing (instead of a flat) dose profile for the modulated beam with a distinctive peak near the end of the range. Radial dose profiles are in reasonable agreement with measurement although GEANT4 produces a sharper penumbra. However, daily variations in the beam output and the type of detector have a significant effect on the measured data.

E.2. Introduction

The Douglas Cyclotron within the Clatterbridge Centre for Oncology (CCO) is the only proton therapy facility in the United Kingdom, and has an energy suitable for treating various forms of eye cancer. There are presently over a dozen centres in the world offering this type of treatment. One of the attractive characteristics of a proton beam is the narrow Bragg peak, which can be spread out uniformly over a certain depth using range modulation. This results in the tumour receiving a homogeneous dose, with negligible dose beyond the proton range.

Many accurate Monte Carlo methods and techniques have been developed for use in conventional radiotherapy involving X-rays, γ -rays, and electrons [0, 0]. This is not, however, the case in proton therapy where the physics of proton dosimetry remains relatively unexplored. Current treatments are predominantly based on calculations using one-dimensional pencil beam algorithms that are assumed to be sufficiently accurate for most purposes [0]. However, the presence of strong inhomogeneities, such as within the head and neck regions, tend to have a significant effect on the accuracy of these simulations

and therefore the treatments that are based on them. More complex pencil-beam algorithms [0] and a Monte Carlo based treatment planning system for proton beams [0] are currently in development.

In this paper, the GEANT4 Monte Carlo code [0] will be used to resolve some of these dosimetry issues in low-energy proton beams. Depth dose characteristics (peak to entrance ratio, Bragg peak width) and radial dose characteristics (rms lateral deflection for pencil beams) as well as distributions of proton energy are calculated using GEANT4 for a number of mono-energetic proton beam energies. All these will then be compared with calculations performed in other work with the following three Monte Carlo codes: PTRAN [0], a system of codes for calculating the transport of protons with initial energies between 50 and 250 MeV in homogeneous water, MCNPX (version 2.5d [beta release]) [0, 0], a general purpose N-particle code similar to GEANT4, and SRIM (version 2003.12) [0], a group of programs for determining the stopping power and range of ions in matter.

The more difficult challenge of simulating range-modulated and full-energy beams from the CCO proton therapy facility, that is similar to most other low-energy clinical proton beam lines, is also described. In particular, depth-dose and radial-dose distributions for these beams calculated using GEANT4, MCNPX and McPTRAN.RZ are compared with diode and film measurement. McPTRAN.RZ is a derivative of PTRAN developed by Palmans [0] for simulating proton beams in a cylindrical-slab geometry. In all cases, the possible reasons for any observed discrepancies between the different codes and the measurements will be discussed.

E.3. Method of calculation

E.3.1. Description of the codes

A new GEANT4 code WaterPhantom was developed to calculate the depth-dose and radial-dose distributions for a series of proton beams incident on the front face of a phantom. Depth-dose was obtained by scoring the total energy deposited in a series of 200 cylindrical slabs of varying thickness and radius centred on the beam axis, and, radial-dose was obtained from the total energy deposited in a series of 100 annular rings of varying width placed at a given depth in the phantom. WaterPhantom also scores, and stores in an external data file, the phase space data (that is, the energy, the position coordinates and the direction cosines) of all particles striking a scoring plane placed at a specified depth in the water phantom. The required energy and angular distributions at a given depth in the phantom were obtained by analysing the raw phase space data using a MATLAB script.

The 62 MeV proton beam line at the CCO described in detail by Bonnett [0] was modelled using a modified version of HadronTherapy, a code supplied as one of the advanced examples with the GEANT4 distribution. All components of the beam line except for the range shifter have been included in the model. In particular, the PMMA modulator wheel that consists of four spokes each with 32 individual slabs was defined precisely giving eight modulator wedges. This can be rotated between 0 and 45 degrees in arbitrary steps using a dedicated messenger class in GEANT4. A final collimator diameter of 30 mm giving a circular beam in the phantom was used in these calculations.

The calculations described in this paper were carried out using GEANT4 release 6.2 (with patch02), the CLHEP version 1.8.1.0 libraries and with the Low Energy Electromagnetic Physics package G4EMLOW version 2.3. The code was installed and compiled on a linux based server and calculations performed on a number of linux computer nodes.

E.3.2. Physics processes and transport parameters

The Low Energy Electromagnetic Physics package [0] extends the validity of electromagnetic interactions in GEANT4 for a variety of particles to the lower energies that are of particular interest in this work. In particular, the package offers a number of alternative parameterisations to model proton energy loss due to electromagnetic interactions at these energies. In this work, the *ICRU_R49p* parameterisation that is based on ICRU Report 49 [0] was used throughout. However, the ‘chemical effect’ [0] that becomes significant for ionisation losses of low energy protons below a few MeV, and also implemented in GEANT4, was not used here as it was found to have no significant effect on the calculated depth dose distributions in water at 50 MeV.

GEANT4 also offers a number of hadronic shower models for simulating hadronic interactions in matter. The low-energy and high-energy parameterised driven models (LHEP and HEP) were used in this work to model all hadronic processes relating to elastic and inelastic interactions with the nucleus. Both the low energy electromagnetic physics package and the LHEP and HEP models are generally recommended for use in medical applications of GEANT4.

A cut value of 0.02 mm was used in the phantom region (using the cut per region feature) and a cut value of 10 mm in all other regions. The former cut value is at least an order of magnitude smaller than the smallest region present in the phantom geometry to ensure that energy deposition is determined as accurately as possible and that any dependencies of the particle transport on the geometry are minimized.

E.3.3. Calculations

Depth-dose calculations were carried out using WaterPhantom for 50, 150 and 250 MeV monoenergetic pencil beams using slab widths of 0.2, 1.0 and 2.0 mm respectively. Energy, angular and radial dose distributions were determined in 10, 30 and 60 mm radius regions for the same energies at depths of $0.5 \times r_0$ and $0.9 \times r_0$ in the phantom, where r_0 is the proton CSDA range for the given energy.

Simulations of the CCO beam line were performed using a monoenergetic pencil beam of protons with an incident energy of 62.5 MeV (which gives a beam energy of approximately 60 MeV at the front face of the phantom). Depth-dose distributions were scored in a PMMA phantom for both a full-energy (that is, no modulator wheel) and a modulated beam using a slab width of 0.2 mm and a circular beam with a diameter of 30 mm (defined by the final collimator). Radial dose distributions were also determined in 20mm radius regions at the front face and at various depths in the phantom.

E.4. Results and discussion

The depth dose curves in water for 50, 150 and 250 MeV monoenergetic pencil proton beams calculated in this work using GEANT4 are compared with similar curves obtained using PTRAN, MCNPX and SRIM in Figures 1-3. Depth-ionisation data (instead of depth dose) has been plotted for SRIM normalized to the PTRAN entrance dose value. The correction required to convert from ionisation to dose in water is assumed to be small and has not been applied here.

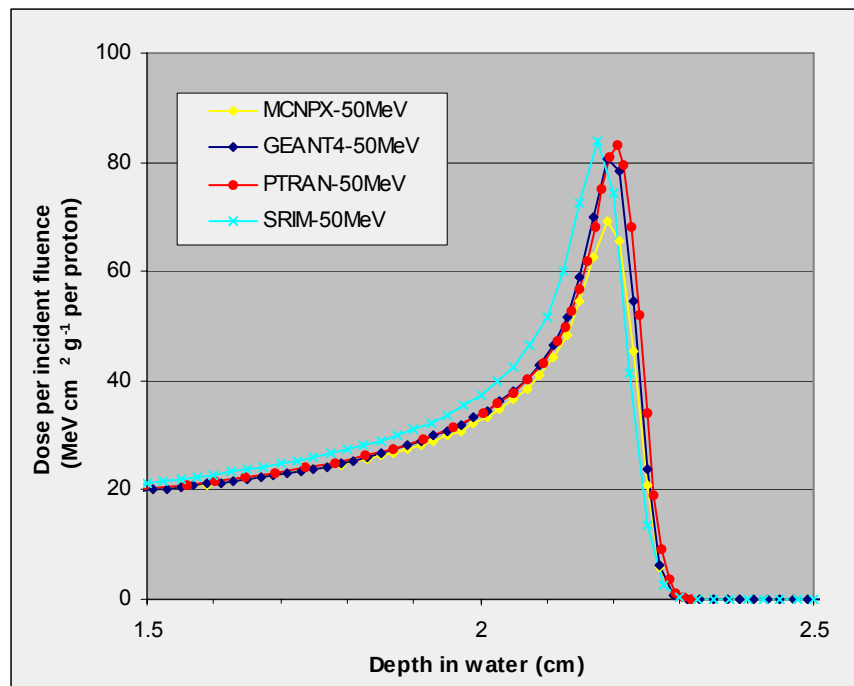


Figure E.1. Depth dose curves in water for a 50 MeV monoenergetic pencil beam.

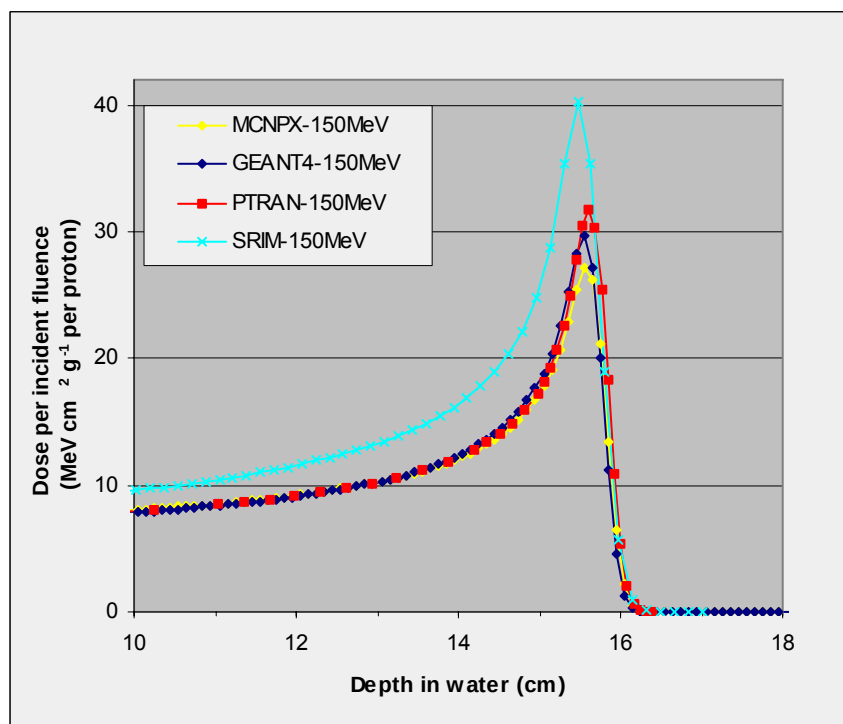


Figure E.2. Depth dose curves in water for a 150 MeV monoenergetic pencil beam.

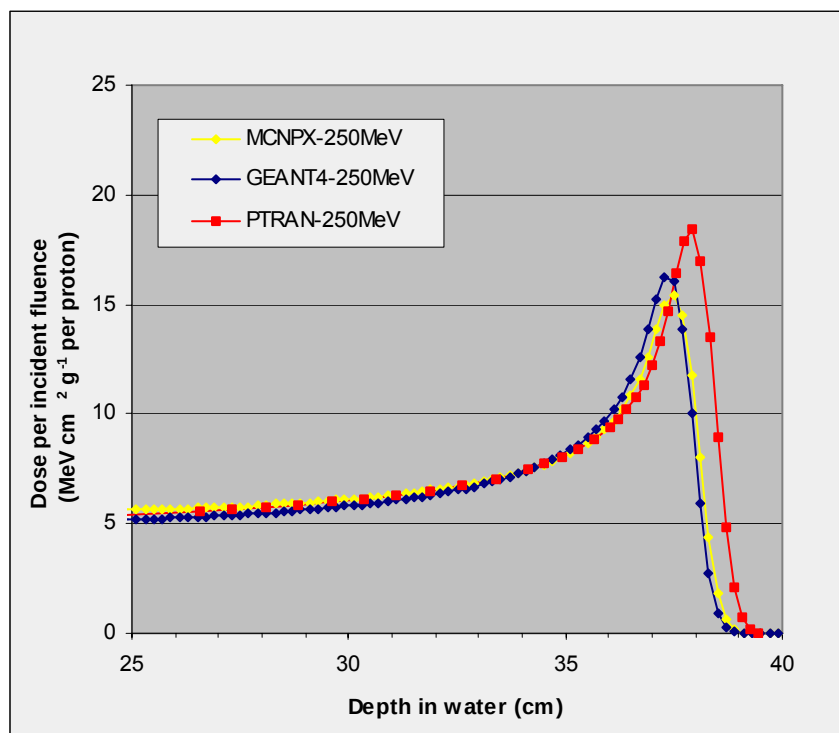


Figure E.3. Depth dose curves in water for a 250 MeV monoenergetic pencil beam.

At 50 MeV, it can be seen that there is good agreement between GEANT4 and PTRAN in both the peak-to-entrance dose ratio (PEDR) and the depth of the Bragg peak. However, there is clearly a range difference between these two codes and SRIM, which has a similar PEDR but produces a Bragg peak at a smaller depth. With MCNPX, the range of the protons is similar to GEANT4 and PTRAN but the PEDR is around 15% smaller. In all cases, the width of the Bragg peak (at 50% of the peak dose) is similar for all codes.

The variation in the range of the protons is primarily due to different proton stopping power data being used in the simulations. Both MCNPX and PTRAN use Janni's shell corrections [0] to stopping powers for protons given in ICRU 49 [0]. GEANT4 uses a parameterised stopping power model also based on ICRU 49 (and, by default, derives the mean excitation energy of the material using Bragg's rule), whereas SRIM uses a stopping power model based on the formulism by Zeigler [0].

The proton energy loss straggling model used in the codes affects both the width and height of the Bragg peak. Both PTRAN and MCNPX use the Vavilov energy straggling distribution [0] whereas GEANT4 uses three different straggling models to simulate energy loss fluctuations [0] including one based on the Vavilov distribution. Nuclear interactions of the primary protons lead to a reduction in the number of protons at depth, and so also in the absorbed dose. This reduces the height of the Bragg peak, but at 50 MeV, only around 5% of the proton energy is lost by this process and the effect is small. However, the Bragg peak on the MCNPX curve is around 20% smaller than the GEANT4 and PTRAN indicating differences in the implementation of the energy straggling model or the use of a different multiple scattering model. The SRIM curve is a few percent higher than the other curves at all depths up to the Bragg peak. This is consistent with the fact that SRIM does not model proton energy loss by nuclear interactions and so instead this energy loss contributes to the calculated ionisation.

At 150 MeV, shown in Figure 2, the agreement between GEANT4 and PTRAN depth dose curves is not as close as at 50 MeV. In particular, the Bragg peak obtained with both GEANT4 and MCNPX now occurs at a slightly smaller depth compared to PTRAN. The PEDR for GEANT4 and MCNPX is now 10 % and 18% smaller respectively than PTRAN. Even though all these codes use similar stopping power models, the implementation and evaluation of these models is different. For example, GEANT4 uses a parameterised fit to the model whereas PTRAN uses lookup tables for proton energy loss based directly on ICRU 49.

Energy losses due to nuclear interactions are much more significant at 150 MeV and 250 MeV (contributing about 20% and 40% of the total energy loss respectively). Consequently, the SRIM curve is now up to 20% higher than the other curves at 150 MeV for the reasons outlined above and so the SRIM results obtained for 250 MeV are not presented. Energy losses at all depths up to the beginning of the Bragg peak are very similar for the other codes at 150 MeV indicating that both the models and nuclear interaction cross-sections used are very similar.

At 250 MeV, shown in Figure 3, the difference in the stopping power data evaluation in the PTRAN compared to GEANT4 and MCNPX has an even more pronounced effect on the depth of the Bragg peak. The discrepancies in the non-elastic nuclear cross sections used in all these codes, indicated by the variation in the height of the Bragg peak and differences in the energy losses at smaller depths, is also more apparent at this energy.

Figures 4 and 5 show the radial dose distributions at depths of $0.5 \times r_0$ and $0.9 \times r_0$ in water for the three proton energies calculated with PTRAN, GEANT4 and MCNPX, where the CSDA range r_0 has been taken from ICRU 49. There seems to be generally good agreement between PTRAN and GEANT4 over small radii at both depths indicating that these codes use similar multiple scattering distributions. However, the tails of the distributions obtained with GEANT4 extend to much greater radii when compared to PTRAN. This is primarily because PTRAN does not transport secondary protons, unlike MCNPX and GEANT4. The radial dose distributions obtained for MCNPX at both depths are generally broader compared to GEANT4 and PTRAN indicating differences in the multiple scattering models being used in this code. However, the tails of these distributions are similar to GEANT4 at all incident energies indicating that the models are more closely matched for lower energy protons. By default, PTRAN uses the Molière formalism [0] to model multiple scattering of charged particles whereas MCNPX uses a model based on Goudsmit and Saunderson [0]. GEANT4 uses a more complete model for multiple scattering based on the theory of Lewis [0].

Figures 6 and 7 show the energy distributions at depths of $0.5 \times r_0$ and $0.9 \times r_0$ in water. Variations in the peak energy from one code to another at $0.9 \times r_0$ are consistent with the variations in the depth of the Bragg peak noted above. The use of a finite sized slab instead of a plane to score particles in GEANT4 is the most likely reason why the peak at 50 MeV occurs at a smaller energy than the other two codes at 50 MeV, as the position of the peak will be very sensitive to the size of the slab used near the Bragg peak. At $0.5 \times r_0$, the energy distributions obtained with the three codes are in good agreement around the peak energy. The low energy tail in the energy spectra at both depths represent the contribution from secondary protons with MCNPX generally producing a slightly larger tail at lower energies than GEANT4: the data from PTRAN have no such tail, since secondary protons are not transported by this code.

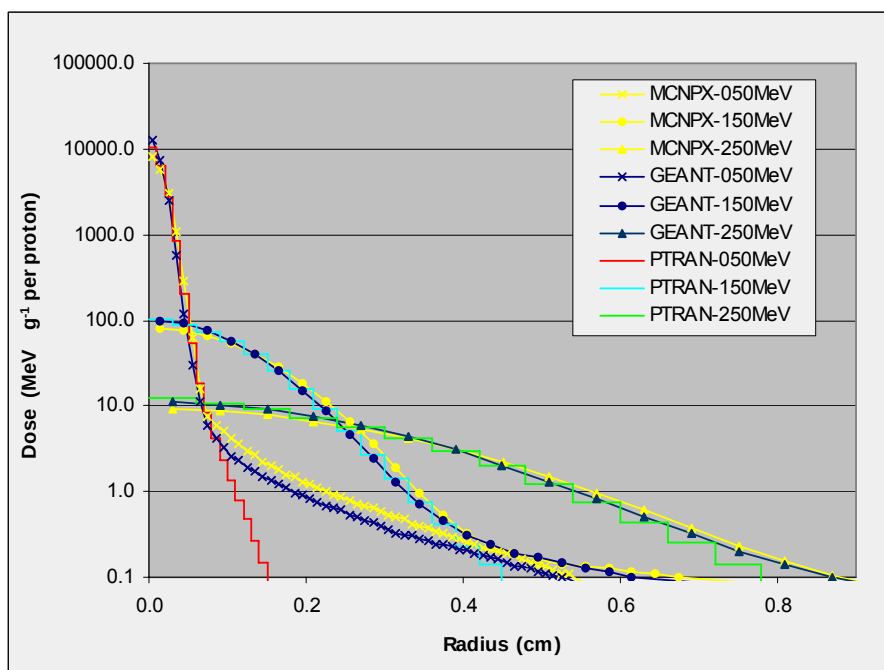


Figure E.4. Radial dose distributions in water at $0.5'r_0$ for monoenergetic pencil beams.

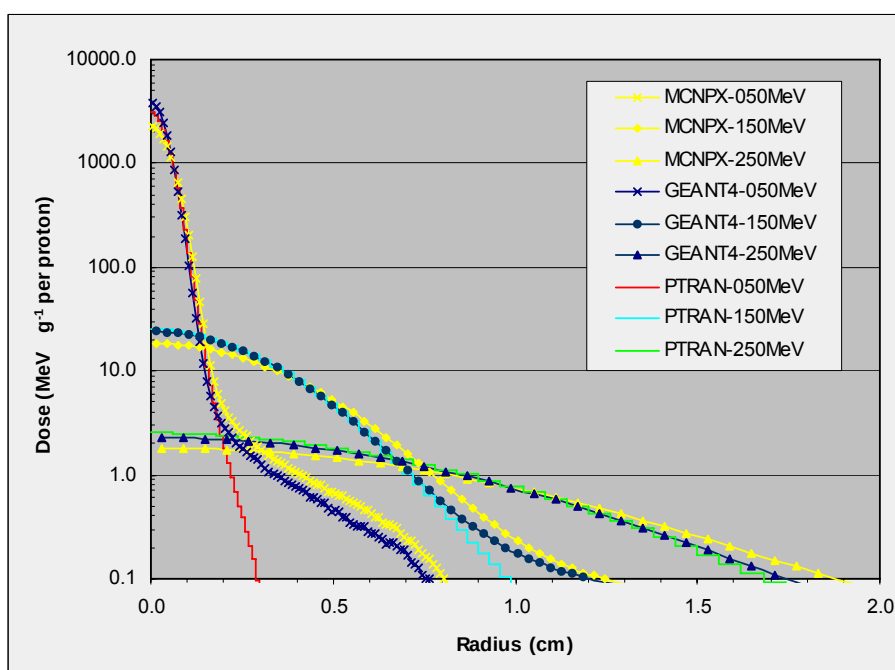


Figure E.5. Radial dose distributions in water at $0.9'r_0$ for monoenergetic pencil beams.

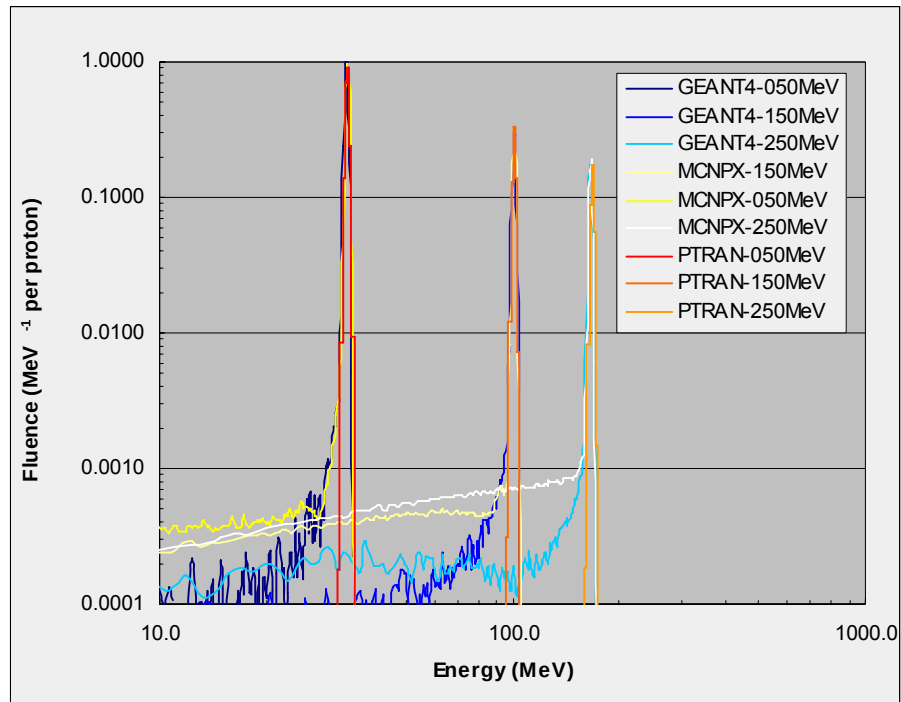


Figure E.6. Energy distributions in water at $0.5 \times r_0$ for monoenergetic pencil beams.

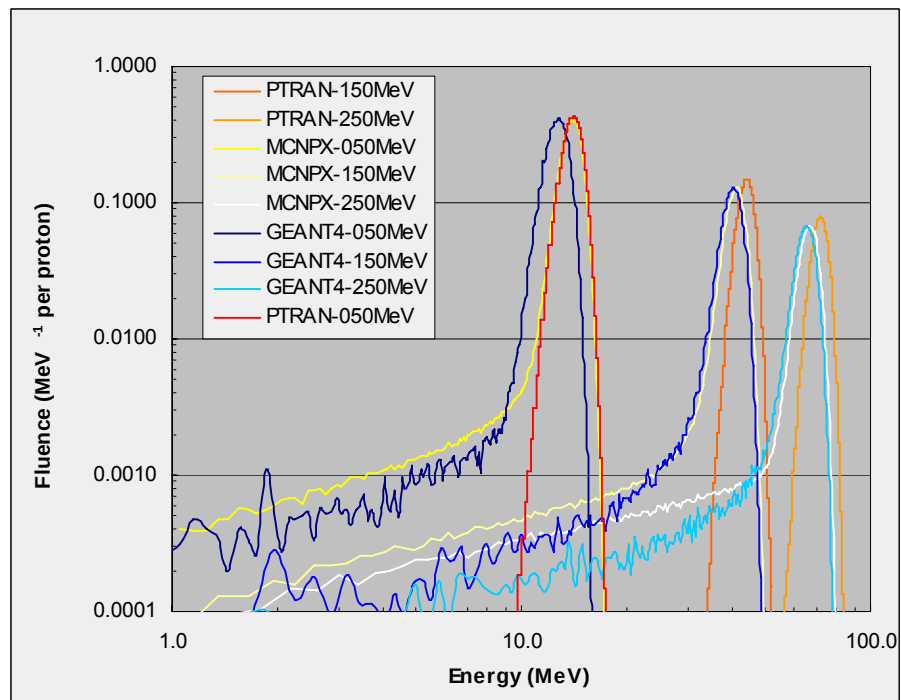


Figure E.7. Energy distributions in water at $0.9 \times r_0$ for monoenergetic pencil beams.

The depth dose curves calculated with GEANT4 in a PMMA phantom for both the full-energy and modulated CCO beam with an initial proton beam energy of 62.5 MeV are compared with recent diode measurements [0] in Figures 8 and 9 respectively. Measurements were made in PMMA but no stopping power correction has been applied to

this data: each was normalised to the entrance dose. Depth dose curves calculated in PMMA with McPTRAN.RZ and MCNPX are also shown. The initial proton energy of 62.5 MeV was obtained by finely tuning the MCNPX simulation so that the average energy at the exit of the final collimator was 60 MeV.

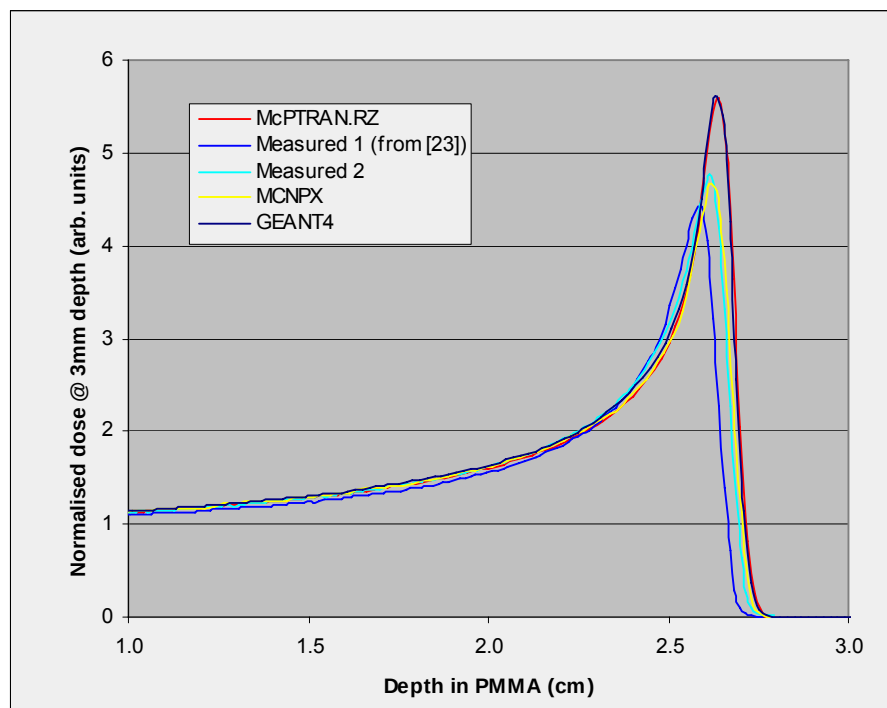


Figure E.8. Depth dose in PMMA for the full-energy (no modulator wheel) CCO beam compared with diode measurements. All curves normalised at the entrance dose.

For the full-energy beam, again it can be seen in Figure 8 that the GEANT4 and McPTRAN.RZ curves in PMMA are very similar, as expected from the earlier observations in a 50 MeV monoenergetic pencil beam, but overestimate the height of the Bragg peak by 20% when compared with measurement. The MCNPX curve, however, is in better agreement with both sets of measurements matching both the depth and the height of the Bragg peak more closely. The widths of the Bragg peaks are similar in all cases. One should note, however, that the CCO beam output can vary significantly. For example, it can clearly be seen in Figure 8 that variations in both the depth and height of the Bragg peak occur for similar diode measurements performed on different days. The type of detector employed for the measurement of dose distributions can also give different results. Silicon diodes tend to give higher response (typically 7% around the Bragg peak) compared to similar measurements with a parallel plate ionisation chamber [0].

For the modulated beam shown in Figure 9, GEANT4 and McPTRAN.RZ both give a larger relative dose with depth in the phantom compared with measurement and, in particular, produce more distinctive peaks in the dose profile near the end of the range. MCNPX produces a generally flatter dose profile over all depths but with a slightly lower relative dose towards the end of the range. In all cases, the dose profiles near the end of the range are particularly sensitive to the construction around the openings between the fins of the modulator wheel. Only minor adjustments to size of the gap and wheel thickness near these openings was found to have a significant effect on the dose profile. The observed

discrepancies in the codes could therefore be partially due to minor differences in the geometry and techniques used to simulate the modulator wheel but they are more likely due to the differences in the physical models used in the codes described previously.

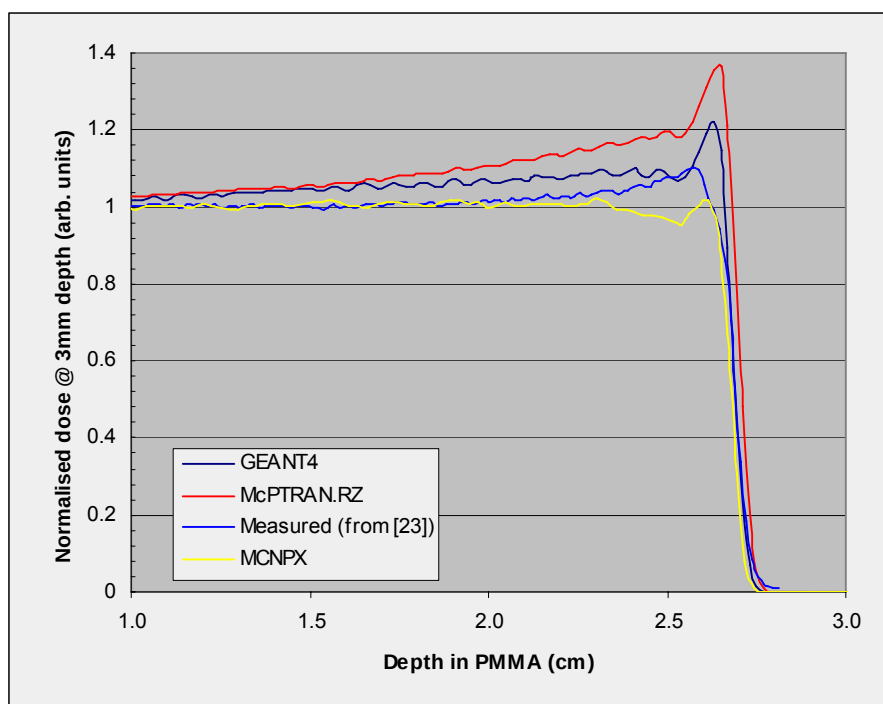


Figure E.9. Depth dose in PMMA for the modulated CCO beam compared with diode measurements. All curves normalised at the entrance dose.

Figures 10 and 11 show the radial dose distributions calculated with GEANT4 for the modulated CCO beam at the front face and at $0.5 \times r_0$ in a PMMA phantom respectively, and compared with both with diode and film measurements (normalised to give the best fit to the calculated data). As mentioned earlier, variations in the measurements tend to be quite large and so the data shown here are obtained from the mean of at least four independent sets of diode measurements and by combining several film profiles both along different axes and from each side of the central beam axis. At the front face, GEANT4 generally gives good agreement with measurement over most of the radial profile up to around 1.2cm. However, there is a distinctive peak towards the edge of the profile compared with measurement. This aspect of the profile near the penumbra is particularly sensitive to the configuration of the final aperture in the beam line that defines the beam size. Although not shown in the measurement data presented here, other measured dose profiles have indicated the presence of these distinctive peaks particularly when close to the final collimator.

The penumbra obtained with GEANT4 is also much sharper than measurement, as shown by the steeper curve at the edge of the profile in Figure 10, but the overall width of the profile is very similar to the film measurement. The diode measurements give a slightly wider profile than film measurement but this may be entirely due to the 1 mm resolution of these measurements with only two measurement points being made in the penumbra region. One should also note that the simulation does not include the cross-hairs that are used to align the beam. This is also likely to have a small effect on the front face profile particularly near the centre of the beam.

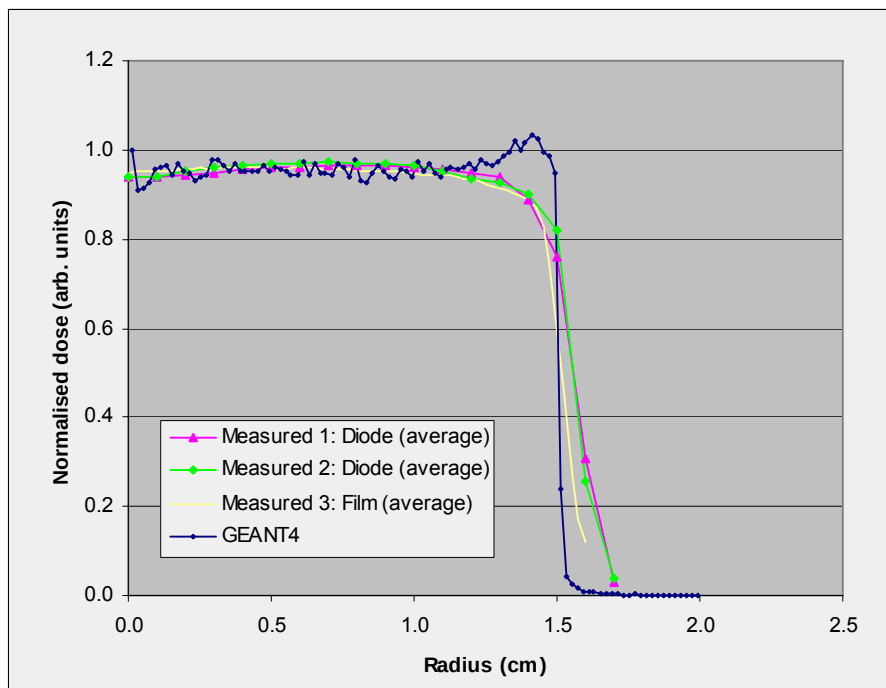


Figure E.10. Radial dose distributions at the front face of the phantom for the modulated CCO beam compared with film and diode measurements.

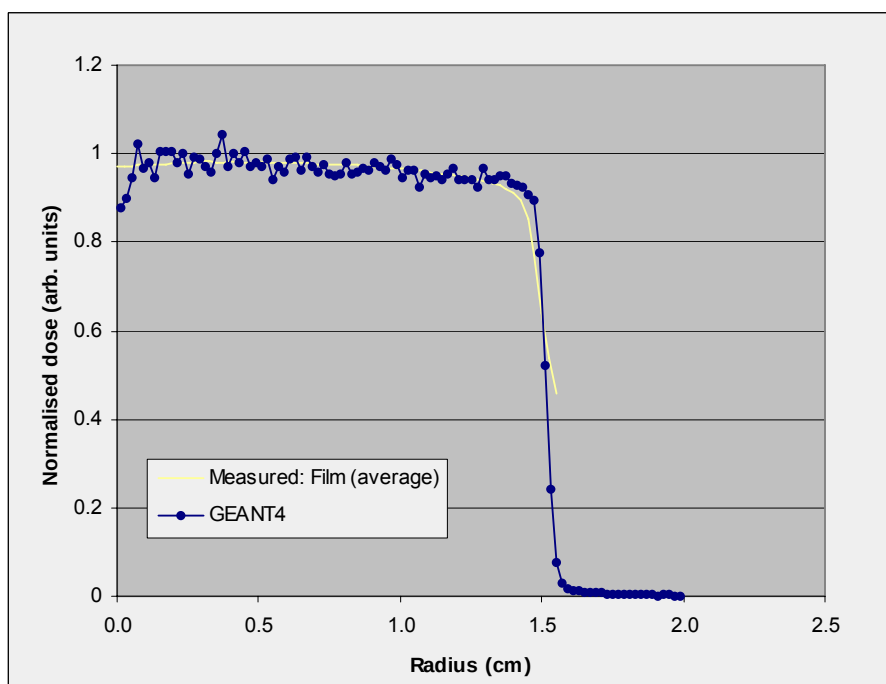


Figure E.11. Radial dose distributions at 0.5'r0 in PMMA for the modulated CCO beam compared with film measurement.

The radial profile obtained at $0.5 \times r_0$ with GEANT4 and shown in Figure 11 is also in generally good agreement with film measurement but with GEANT4 again giving a sharper penumbra. In this case, there is no distinctive peak at the edge of the profile, as seen at the

front face, confirming that the peak is primarily due to low energy proton scatter from the final aperture.

E.5. Conclusions

The work described in this paper has highlighted a number of differences in proton beam simulations with GEANT4, when used with physics list and transport parameters described earlier, and other similar Monte Carlo codes (PTRAN, MCNPX, SRIM). GEANT4 generally agrees with PTRAN in depth dose calculations in water at lower energies (below 150 MeV) but tends to underestimate both the depth and height of the Bragg peak at higher energies primarily due to the method used for deriving the required proton stopping power data and differences in the non-elastic nuclear interaction cross sections. These two codes also agree in their calculation of radial dose and energy distributions around the peak of the distribution with significant differences only occurring at depths towards the end of the range. Unlike GEANT4, PTRAN does not transport secondary protons and so underestimates the tails of these distributions.

GEANT4 is in closer agreement with MCNPX at all energies compared to PTRAN but tends to overestimate the height of the Bragg peak at lower energies. However, the different nuclear interaction cross-sections being used in MCNPX tend to overestimate the dose from these processes at higher energies at depths below the Bragg peak. Similarly, MCNPX gives a broader radial dose distribution than GEANT4 at all energies indicating differences in the multiple scattering models used in these codes. The tails of these distributions are however in closer agreement. The energy distributions of GEANT4 and MCNPX are in generally good agreement at these energies but with MCNPX producing a significantly larger tail at lower energies than GEANT4.

When used to simulate a typical ocular proton beam line, such as the one at the CCO, the depth dose curves calculated with GEANT4 in PMMA for both a full-energy and modulated beam have characteristics that closely match those of measured curves. However, GEANT4 overestimates the height of the Bragg peak for the full-energy beam by around 20% and, tends to give a slightly increasing dose profile (instead of a flat profile) for the modulated beam with a distinctive peak near the end of the range. These observed features are very sensitive to the configuration of the modular wheel, particularly around the gaps and, thus may be partially due to minor differences in the geometry and techniques used to simulate the modulator wheel. However, it is more likely that the different physical models used in the codes account for these differences.

The radial dose profiles obtained with GEANT4 for the modulated beam are also in generally good agreement with measurement. However, GEANT4 produces a sharper penumbra than indicated by measurement and the edge of radial dose profile near the surface of the phantom is sensitive to the configuration of the hole in the final collimator. One should note that all measurements are subject to daily variations in the beam output and have some dependence on the type of detector used.

E.6. References

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ADDENDUM F: PALMANS ET AL (2006)

PALMANS, H., THOMAS, R. and KACPEREK, A. Recombination correction in the CCO low-energy proton beam. *Phys. Med. Biol.*, 2006, **51**, 903-917.

F.1. Abstract

Most codes of practice for dosimetry of proton beams do not give a clear recommendation on the determination of recombination correction factors for ionisation chambers.

In this work, recombination corrections were measured in the low-energy clinical proton beam of the Clatterbridge Centre of Oncology (CCO) using data collected at different dose rates and different polarising voltages. This approach allows the separation of contributions from initial and volume recombination and was compared with results from extrapolation and two-voltage methods. A modified formulation of the method is presented for a modulated beam in which the ionisation current is time dependent. Using a set-up with two identical chambers placed face-to-face yielded highly accurate data for plane-parallel ionisation chambers. This method may also be used for high-energy photon and electron beam dosimetry.

In typical dose rates of 26 Gy min^{-1} used clinically at the CCO, the recombination correction is 0.8% and thus of importance for reference dosimetry. The proton beam should be treated as purely continuous given the high pulse repetition frequency of the cyclotron beam. The results show that the volume recombination parameter for protons is consistent with values measured for photon beams. Initial recombination was found to be independent of beam quality, except for a tendency to increase at the distal edge of the Bragg peak; this is only relevant for depth dose measurements. Using a general equation for recombination and generic values for the initial and volume recombination parameters ($A = 0.25 \text{ V}$ and $m^2 = 3.97 \cdot 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$), the experimental results are reproduced within 0.1% for all conditions met in this work.

For the CCO beam and similar proton beams used for treating optical targets operating at high dose rates, the recombination correction factor can be overestimated by up to 2%, resulting in an overestimation of dose to water by the same amount, if the recommendation from IAEA TRS-398, which is only valid for pulsed beams, is followed without consideration.

F.2. Introduction

Ionisation chamber dosimetry is the most widespread method for reference dosimetry in clinical proton beams. One of the corrections that needs to be applied to the ionisation measurement is for the loss of ions due to recombination. Whilst this correction has been studied thoroughly for high-energy photon and electron beams, for proton beam dosimetry, recommendations in the codes of practice for dosimetry remain vague.

Several physical processes can lead to recombination, but the ones that are commonly considered important are initial recombination and volume recombination (Boag 1966). Initial recombination, taking place within the initial ionisation track, is assumed to be independent of the ionisation current but it is dependent on the ionisation density along the track and thus on the beam quality. In heavy ions for example it is shown to vary considerably with beam quality (Kanai et al. 1998). Volume recombination, which is due to recombination of charges from different tracks, depends on the ionisation current and hence

dose rate. Codes of practice for dosimetry usually give recommendations for the determination of recombination, the most common method being the two-voltage technique. Continuous and pulsed beams are treated differently in most dosimetry protocols. Boag (1996) has shown that for continuous beams, the inverse of the current varies linearly with the inverse square of the polarising voltage, whereas for pulsed radiation it varies, in first order, linearly with the inverse of the polarising voltage. For the Boag theory for pulsed beams to be applicable the length of the pulse should be very short compared to the ion collection time and the time between pulses should be much larger than the collection time.

In clinical proton beam dosimetry, the recombination correction of ionisation chamber currents is usually treated in the same way as for high-energy photon and electron beams. In AAPM report 16 (Lyman et al. 1986) it is recognised that the conditions for pulsed beams are usually not fulfilled. Although proton beams accelerated by an isochronous cyclotron are pulsed beams for example, the pulse repetition rate is so high (of the order of 20 MHz) that the second condition is not fulfilled (the ion collection time is about 0.25 ms for a polarising voltage of 100 V and a plate separation of 2 mm). In fact, the displacement of ions during the time between two pulses is small and consequently such proton beams ought to be treated as continuous beams. For modulated⁷ proton beams, the time dependent dose could be well regarded as pulsed due to the modulation process. However, even for the deepest points on the spread-out Bragg peak, the duration of each pulse is usually long compared to the ion collection time and again the beam should be regarded as continuous in relation to recombination.

AAPM report 16 gives some detail on how to treat initial and volume recombination and recommends plotting ‘the reciprocal of the observed ionisation current against an appropriate function of the collecting field strength’. Other codes of practice like ECHED (Vynckier et al. 1991, 1994) and ICRU report 59 (1998) give little detail about how to determine the correction for recombination. IAEA TRS-398 notes that some proton beams should be regarded as continuous but this is only made explicit for ‘proton synchrotron beams of long pulse duration and low pulse repetition frequency’. Unfortunately, only the approach for pulsed beams is retained in the rest of the code of practice for proton beam dosimetry and is the only option considered in the worksheets.

In the experimental literature on ionisation chamber dosimetry of protons or heavy ions, recombination correction factors are often ignored or else the two-voltage technique is used assuming continuous or pulsed beams, and the recombination corrections are usually found to be small at the low dose rates used in these experiments (Siebers et al. 1995, Vatnitsky et al. 1996, Palmans et al. 1996, Hartmann et al. 1999, Moyers et al. 2000 and Palmans et al. 2002). Some experiments investigated recombination curves by measuring at a series of voltages in continuous radiation in an isochronous cyclotron beam (Palmans et al. 2001) (separating initial and volume recombination) or in pulsed radiation of a synchrotron beam (Nothomi et al. 2001).

The dose rate used for treatments at the 62 MeV proton beam of the Clatterbridge Centre of Oncology (CCO) is typically 26 Gy min⁻¹ and it can be expected that the recombination correction is substantial. In this paper, we measure the recombination correction factor for NACP02, Roos, Markus and NE2561 type chambers in the CCO beam at various dose rates, which allows the separation of the initial recombination and volume recombination contributions. The aim is to compare the initial and volume recombination parameters with those from high-energy photon and electron beams, to determine any significant dependence

⁷ The term modulated beam here and further in this paper is used to denote a spread out Bragg peak (SOBP) exhibiting a constant dose over a substantial depth (in this work from the surface to the distal edge).

of the initial recombination on beam quality and to verify the magnitude of the deviations with the conventional two-voltage method .

F.3. Method

F.3.1. Measurement of initial and volume recombination

The ratio of the saturation current and current at operating voltage V for continuous beams can be approximated by (see e.g. Boutillon 1998):

$$\frac{I_{sat}}{I_V} = 1 + \frac{A}{V} + \frac{m^2 g}{V^2} \cdot I_{sat} \quad (F.1)$$

where m^2 is the volume recombination parameter which is a property of the gas in the cavity and which depends on the recombination coefficient and the mobility of negative and positive ions in the gas. The second term on the right hand side represents initial recombination and the third term represents volume recombination. A and g depend on chamber geometry. For a plane-parallel chamber g equals $d^4/6v$, where d is the plate separation and v the collecting volume of the chamber. The plate separation is nominally 2 mm for all the chambers used in this study. The volume v can be estimated as the nominal volume for the ionisation chamber type or it can be derived from the calibration factor of the ionisation chamber as

$$v = \frac{(W_{air}/e) \cdot s_{w,air} \cdot p}{\rho \cdot N_{D,w}} \quad (F.2)$$

where (W_{air}/e) is the mean energy required to produce an ion pair in dry air by electrons in the ^{60}Co calibration beam, $s_{w,air}$ is the water to air stopping power ratio in ^{60}Co , p is the ion chamber perturbation correction factor for the ionisation chamber in ^{60}Co , ρ is the density of air at the calibration reference conditions and $N_{D,w}$ is the absorbed dose to water calibration factor in ^{60}Co of the ionisation chamber. The numerical values of (W_{air}/e) , $s_{w,air}$ and p were taken from IAEA TRS-398 and the calibration factors $N_{D,w}$ were determined at NPL.

For a cylindrical chamber the expression for g becomes more complicated. According to Boag (1996), an equivalent electrode spacing should be taken into account. Furthermore, most cylindrical chambers are in reality thimble chambers and thus the recombination can be different in the cylindrical part and the hemispherical part. Basically, d in the expression for g should be replaced with $(a-b)*K$ where a is the radius of the cavity, b is the radius of the electrode and

$$K = \sqrt{\frac{1}{2} \cdot \frac{a+b}{a-b} \cdot \ln(a/b)} \quad (F.3)$$

for a cylindrical geometry, and

$$K = \sqrt{\frac{1}{3} \cdot \left(\frac{a}{b} + 1 + \frac{b}{a} \right)} \quad (F.4)$$

for a spherical geometry.

g for the thimble chamber can then be derived from g_{cyl} for the cylindrical part and g_{spher} for the spherical part as:

$$g = \frac{v_{cyl}^2 \cdot g_{cyl} + v_{spher}^2 \cdot g_{spher}}{(v_{cyl} + v_{spher})^2} \quad (F.5)$$

where v_{cyl} and v_{spher} are the volumes of the cylindrical and hemispherical part of the chamber.

It is interesting to investigate what can be expected from different approaches to evaluate the recombination correction for a virtual beam that behaves as described in equation (F.1). This approach will be compared with the experimental results. It will also be compared with the theoretical result for a pulsed beam (Boag 1966), which says that in the first order, the ratio I_{sat}/I_V is linearly dependent on $1/V$.

For a $1/n$ fraction of a polarising voltage V , where n is larger than unity but not necessary an integer, equation (F.1) becomes:

$$\frac{I_{sat}}{I_{V/n}} = 1 + n \cdot \frac{A}{V} + n^2 \cdot \frac{m^2 g}{V^2} \cdot I_{sat} \quad (F.6)$$

Dividing equation (F.6) by equation (F.1), neglecting higher order terms, and approximating I_{sat} with I_V in the third term:

$$\frac{I_V}{I_{V/n}} \approx 1 + (n-1) \cdot \frac{A}{V} + (n^2-1) \cdot \frac{m^2 g}{V^2} \cdot I_V \quad (F.7)$$

The approximation is thus only valid if I_V is close to I_{sat} . More details on this derivation are given by Boutillon (1998). Measuring this ratio for a range of currents I_V and a range of voltages gives a more accurate determination of both terms than can be obtained from an extrapolation to $1/V = 0$ or from the two-voltage method (which is not able to separate the terms at all).

As indicated in the introduction, the modulated beam should be regarded as continuous. For the deepest points on the SOBP, where only one modulation contributes to the signal, there might be a concern about the continuous character of the beam since the length of the pulse is only a factor of eight longer than the ion collection time. There will be a fraction of the time interval in which the charge distribution in the cavity is built up and after the pulse, there will be a sweep of the existing equilibrium charge distribution which should be treated in a similar way to how a pulse is swept away in Boag's theory for pulsed beams (Boag 1966). In this work we have assumed that the contribution of this transition effect on the recombination is negligible.

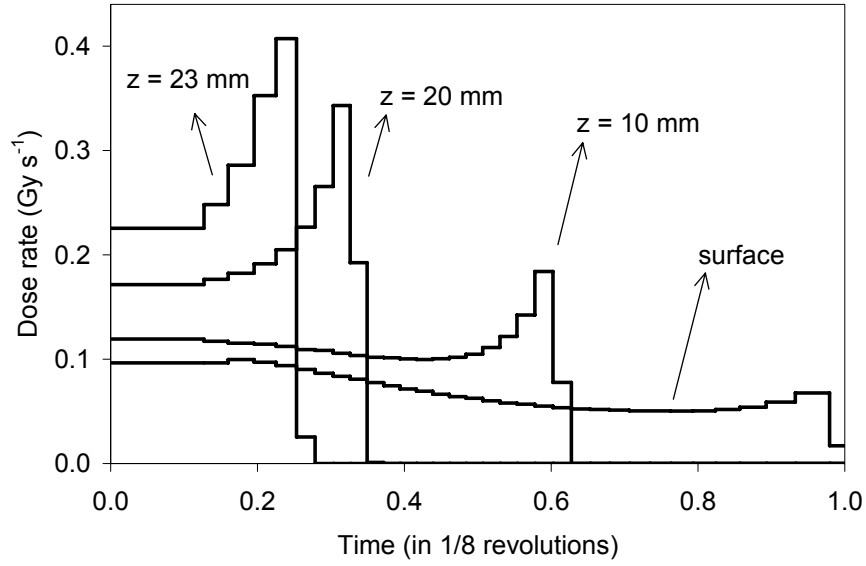


Figure F.2. Ionisation current as a function of time, at various depths on the spread-out Bragg peak.

For a modulated beam we need to account for the fact that the dose rate is a function of time and that this time dependent function varies with depth. This is shown in figure F.2 where the dose rate at various depths is plotted against time for the modulated CCO beam used in this work. The entire range of the horizontal axis corresponds to 1/8 revolution of the modulating wheel or to about 0.01s. The dose rate for each wheel step was derived from the measured depth dose rate in the non-modulated beam. To this end, the dose rate at a depth corresponding with the thickness of the wheel + the depth in the phantom was corrected for the number of protons that are lost from the beam for each thickness of the wheel. This number was determined experimentally during the design phase of the wheel. It would be obvious to characterise the recombination as a function of time averaged dose but a more correct way of analysing recombination is to consider an effective current. It can be proven that, after neglecting higher order terms, the division of equations (F.6) and (F.1) for a modulated beam gives

$$\frac{\langle I_V \rangle}{\langle I_{V/n} \rangle} \approx 1 + (n-1) \cdot \frac{A}{V} + (n^2-1) \cdot \frac{m^2 g}{V^2} \cdot \frac{\langle I_{sat}^2 \rangle}{\langle I_{sat} \rangle} \quad (\text{F.8})$$

where the expression in angle brackets denote time average values, i.e.

$$\langle I_V \rangle = \int I_V \cdot dt / \int dt, \quad \langle I_{sat}^2 \rangle = \int I_{sat}^2 \cdot dt / \int dt, \text{ etc.}$$

Note that a similar approach could be used in any radiation field that results in a time varying ionisation current, such as for example a scanned proton beam, with the only concern that if the peak current becomes very high it might be necessary to take higher order terms into account (this especially occurs when the distal edge of the target is irradiated with the Bragg peak and it becomes very important when intensity modulation is used while sweeping with a constant velocity). The last factor in the third term of equation (F.8) can be regarded as the effective current $I_{V,eff}$, which we can approximately separate in two factors; one that is the time average current $\langle I_V \rangle$ as measured by the integrating electrometer and one that is independent of the dose rate:

$$I_{V,eff} \approx \frac{\langle I_V \rangle}{\langle I_{sat} \rangle} \cdot \frac{\langle I_{sat}^2 \rangle}{\langle I_{sat} \rangle} = \langle I_V \rangle \cdot \frac{\langle I_{sat}^2 \rangle}{\langle I_{sat} \rangle^2} \quad (F.9)$$

This approximation assumes again that I_V is close to I_{sat} . The quotient $\langle I_{sat}^2 \rangle / \langle I_{sat} \rangle^2$ can be derived from the simulated time dependence of the ionisation current (proportional to the time dependent dose rate plotted in figure F.2).

Expression (8) thus becomes:

$$\frac{\langle I_V \rangle}{\langle I_{V/n} \rangle} \approx 1 + (n-1) \cdot \frac{A}{V} + (n^2-1) \cdot \frac{m^2 g}{V^2} \cdot I_{V,eff} \quad (F.10)$$

For both non-modulated and modulated beams, once the resulting parameters A/V and $m^2 g/V^2$ have been determined experimentally, the recombination correction factor can approximately be expressed as a function of the ionisation current of the ionisation chamber as:

$$p_{ion} \approx 1 + \frac{A}{V} + \frac{m^2 g}{V^2} \cdot I_{V(eff)} \quad (F.11)$$

where for the modulated beam, the effective current $I_{V,eff}$ has to be derived from the time average current using equation (F.9).

F.3.2. Proton beam and experimental set-up

The 62 MeV proton beam at CCO is generated by a fixed energy isochronous (25.8 MHz) cyclotron and is used for the treatment of ocular melanoma. A detailed description of the beam line has been given previously (Bonnett et al. 1993). The beam characteristics were the same as those reported by Palmans et al. (2004). In this work, a 30 mm diameter circular brass aperture was used as a final collimator. Monitoring of the beam was performed with a transmission ionisation chamber prior to the collimator. However, as noticed by Palmans et al. (2004) small fluctuations of the beam cause scatter in the measured ratio of the output of the final collimated field and the (larger) integrated field at the position of the transmission monitor. This scatter was at a level that made it impossible to determine the small corrections for recombination accurately. In order to overcome this monitoring problem, for the plane-parallel chamber measurements two identical chambers were positioned face to face in a PMMA phantom as shown in figure F.1a. In this way the proton fluences through both chambers are highly correlated. For the cylindrical chamber, measurements were performed in air with its centre on the beam axis and an NACP02 chamber was positioned, also with its centre on the beam axis, immediately beyond the cylindrical chamber as is shown in figure F.1b. A 2mm PMMA plate was positioned between the collimator and the cylindrical chamber in order to absorb secondary protons from the beam line. Although the geometrical shape of both chambers is different, their effective area perpendicular to the beam axis is similar and thus the degree of correlation was also high. All the measurements in the non-modulated beam were performed close to the phantom surface (without any additional material between the beam and the back of the ionisation chamber IC1 in figure F.1a). In the modulated beam, measurements using one Markus chamber were performed at various depths in order to investigate any possible dependence with beam quality. In order to take this to the extreme, the depth on the distal edge where the signal dropped to 60% of the signal at the centre of the SOBP was found by ‘tuning’ the amount of PMMA in front of the phantom. At that depth, the dose-weighted average LET is about a factor of eight larger than at the surface (see e.g. Gustavsson et al. 2004).

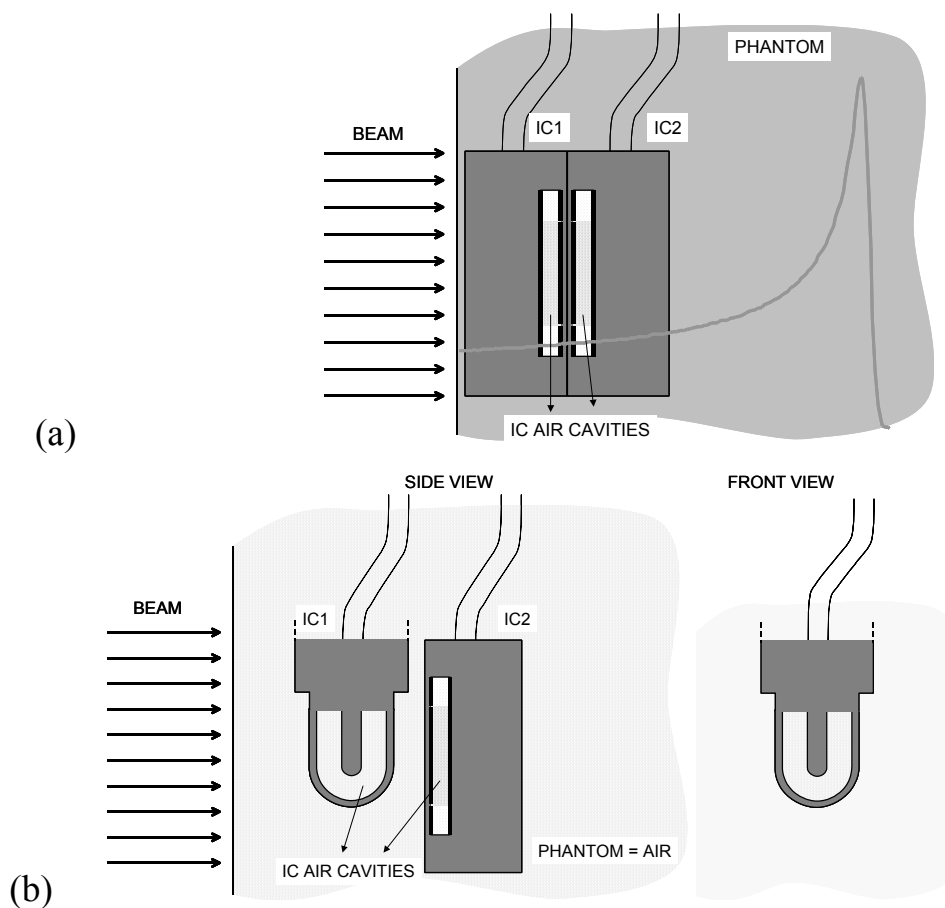


Figure F.1. Set-up of (a) two identical plane parallel chambers and (b) of a cylindrical chamber, with a plane-parallel chamber as monitor.

Measurements were performed at four voltages: the operating voltage V , and $V/2$, $V/3$ and $V/4$. For the plane-parallel chambers the operating voltage was -100 V and for the cylindrical chamber -200 V, which are recommended values for the chamber types used in this study and generally used in the UK for high-energy photon and electron beams. It can be argued that it is not uncommon that these chambers are operated at higher voltages (e.g. Onori et al. 2000, Kacperek et al. 2002, Newhauser et al. 2002). However, it has been demonstrated by for plane-parallel ionisation chambers (e.g. by Burns and McEwen 1998 and McEwen et al. 2001) and for cylindrical ionisation chamber (e.g. by Zankowski and Podgorsak 1998) that the linear models for recombination (such as the Boutillon and Boag models) break down at higher polarising voltages due to charge multiplication. In the connection to our measuring assembly the outer electrode (the thimble wall of the NE2561 cylindrical chamber or the entrance window of the plane parallel chambers) is at the applied polarising voltage and the collecting electrode is close to earth potential. It was verified for one plane-parallel and one cylindrical chamber that the polarity effect is negligible.

Dose rates were not constant over the entire duration of the experiment but were approximately 8 Gy min^{-1} , 20 Gy min^{-1} and 32 Gy min^{-1} at the surface for the non-modulated beam and 6 Gy min^{-1} , 16 Gy min^{-1} and 26 Gy min^{-1} in the centre of the SOBP of the modulated beam.

F.4. Results

Since for each chamber type, both chambers were alternatively used as the chamber under test and as the monitor, we had some concern about the effect of the orientation of the chamber closest to the beam (see figure F.1a). For this reason, one set of measurements was performed with both NACP02 chambers with their front windows oriented towards the beam. The results with this configuration were not significantly different to those obtained using our preferred face-to-face set-up. Another concern was a potential effect of changing the polarising voltage on one chamber to the electric field in the other chamber. For that reason, a set of measurements was performed with a larger spacing between both ionisation chambers, again giving no significant difference. For all the remaining plane-parallel chamber measurements reported here we used the face-to-face configuration with the front windows of the chambers in contact with each other.

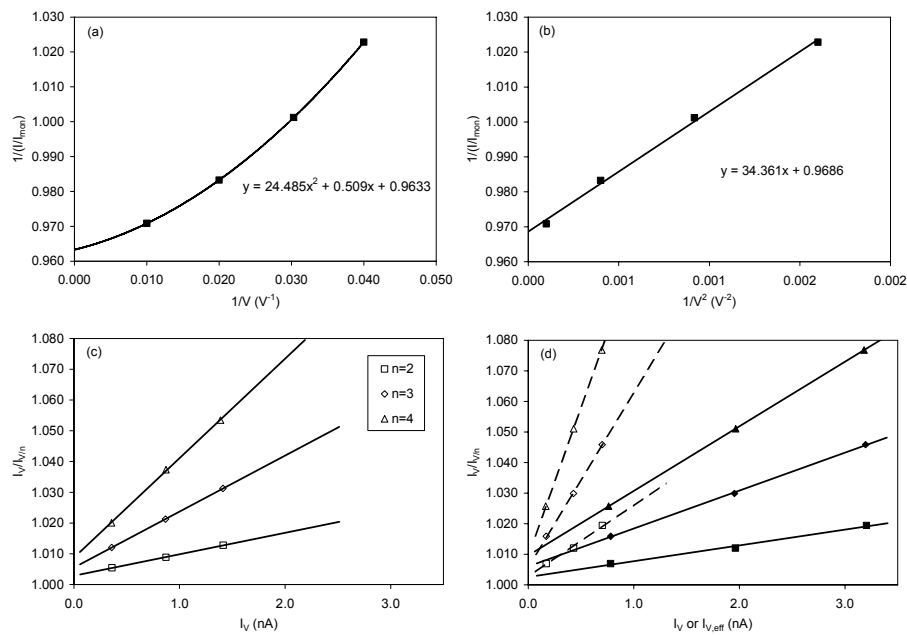


Figure F.3. Experimental data for the Markus SN 478: inverse of the ionisation current at a dose rate of 32 Gy min^{-1} in a non-modulated beam (a) plotted against the inverse of the polarising voltage with a quadratic fit and (b) plotted against the inverse square of the polarising voltage with a linear fit, (c) ratio of I_V and $I_{V/n}$ for three voltage divisions and three ionisation currents I_V with linear fits as a function of I_V in a non-modulated beam and (d) similar data obtained in a modulated beam at a residual range $R_{res} = 0.6 \text{ cm}$. In figure (d) the open symbols with dashed fitted lines represent data as a function of I_V , the time averaged ionisation current, whereas the solid symbols with solid fitted lines are data as a function of $I_{V,eff}$ as defined in equation (F.9).

Figure F.3 presents some of the results obtained for the Markus SN478 with the Markus SN 2225 as the monitor. The results for all other chambers were similar. Figures F.3a and F.3b show the inverse of the measured current (per unit current of the monitor) in the non-modulated proton beam for different polarising voltages. The dose rate at the phantom surface was 32 Gy min^{-1} . The type A uncertainty on the measured data points was 0.01% or

smaller. It is clear that the inverse of the current does not vary linearly with $1/V$ showing that Boag's theory for pulsed beams is not applicable here. Consequently, applying the two-voltage method for any two points on this curve will result in large errors on the determination of p_{ion} . In principle, the parameters A and m^2 in equation (F.1) can be derived from the quadratic fit in figure F.3a, but it is obvious that such an extrapolation would be very sensitive to fluctuation in the data points. Furthermore, it is only due to the extremely low type A (statistical) uncertainties in our measurements that four points are sufficient to make a reasonable fit, which could yield reasonable estimates for the values of the parameters A and m . The extrapolation to $1/V = 0$ results in a recombination correction factor $p_{\text{ion}} = 1.0078$. In figure F.3b, it is also clear that there is a small but, given the small uncertainties on the data points, significant deviation from linearity as a function of $1/V^2$, indicating that the initial recombination term is not negligible.

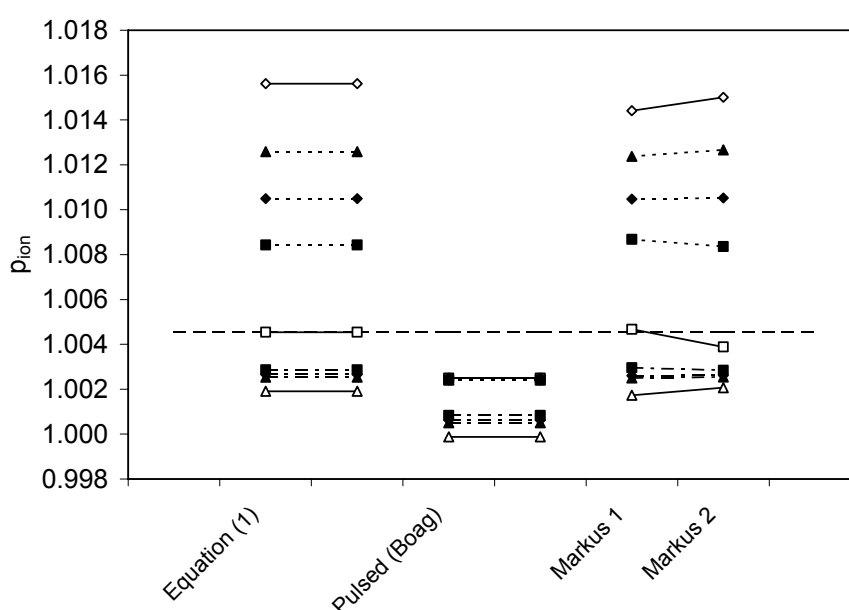


Figure F.4. Recombination correction factors determined using a variety of methods: linear and quadratic fits in a $1/I$ versus $1/V$ plot (open diamonds and open squares respectively), linear fits in a $1/I$ versus $1/V^2$ plot (open triangles), two-voltage method for three voltage divisions using the coefficient expressions for pulsed beams given in IAEA TRS-398 (solid squares, diamonds and triangles connected by dashed lines for $V/2$, $V/3$ and $V/4$ respectively) and two-voltage method for three voltage divisions using the expression for continuous radiation given in IAEA TRS-398 (solid squares, diamonds and triangles connected by dashed-dotted lines for $V/2$, $V/3$ and $V/4$ respectively). These methods were applied to data generated for a virtual beam according the model of equation (F.1) (left), to data generated for a virtual beam according to Boag's model for pulsed beams (centre) and (c) to measurements with two Markus chambers in a non-modulated beam at a dose rate of 20 Gy min^{-1} (right). The numerical values for the parameters in the virtual models are indicated in the text. The dashed horizontal line at $p_{\text{ion}} = 1.0045$ represents the value obtained with equation (F.11).

As indicated earlier, it is useful to compare the results obtained with various methods (fits as in figures F.3a and F.3b or the two-voltage method) to derive p_{ion} from the measured data with the results for a theoretical description of the beam. This way a qualitative assessment can be made to find out if the beam behaves predominantly as a continuous or pulsed beam or as a mixture of both. The result of such an exercise is shown in figure F.4. The experimental results were for two Markus chambers used in this study measured in the non-modulated beam with a dose rate at the phantom surface of 20 Gy min^{-1} . This resulted in an ionisation current of about 0.85 nA . The results were compared with a theoretical model according to equation (F.1). The expected ionisation current as a function of the saturation current and as a function of polarising voltage was calculated using the value of $m^2 = 3.97 \cdot 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$ for X-rays taken from Boutillon (1998) and a value of $A = 0.25 \text{ V}$ which was around the typical values found experimentally for most chambers in this study. Then the same list of methods for evaluating the recombination correction as for the experimental set of data was used. For the results based on a Boag theory compliant pulsed beam, the same value for the initial recombination term was used and the dose per pulse was derived for the same ionisation current and a pulse repetition frequency of 25.8 MHz from the cyclotron to generate a set of expected ionisation currents as function of the saturation current and the polarising voltage.

Figure F.4 clearly illustrates that for all the considered extrapolations the results with the virtual beam according to equation (F.1) are qualitatively in good agreement with the experiments, whereas the virtual pulsed beam yields results that are not. This demonstrates that the ionisation current behaves predominantly as for a continuous beam. A similar exercise for the modulated beam, where the pulse frequency was assumed to be the modulation frequency, predicts similar results for the continuous beam theory but again results that deviate substantially from the predicted results for a pulsed beam. In this case results are much larger than what is found experimentally, illustrating also that the pulses from the modulation are long compared to the ion collection time and that the sweeping effects in the begin and end of the modulation pulse are negligible. Figure F.4 gives also an indication of the magnitude of the errors that can be expected when one would apply the two-voltage method assuming the theory for pulsed beams. If the voltage ratio is 4 then the error in this example is about 0.8% . If the two-voltage expression for continuous beams is used (assuming a linear relation between $1/I$ and $1/V^2$), the recombination correction is in general underestimated because the initial recombination term is not properly dealt with. The error is, however, much smaller. It can be seen in figure F.4 that a quadratic fit in a $1/I$ versus $1/V$ plot gives for the virtual beam a perfect agreement with equation (F.11), whereas for the experimental beam the differences are small. In general, the agreement using this model was reasonable although larger deviations were observed, mainly illustrating that the extrapolation that needs to be made in such a plot is highly sensitive to small fluctuations in the experimental data points. Moreover, the extrapolation is complicated by the limited maximum polarising voltage that can be applied before other effects such as charge multiplication start to distort the $1/I$ versus $1/V^2$ dependence (Zankowski and Podgorsak 1998).

Figure F.3c presents the results for the Markus chamber in the non-modulated beam at three different dose rates from which the parameters A and m , according to equation (F.7), could be derived. Figure F.3d presents similar results for the modulated beam at a residual range⁸ of 0.35 cm , i.e. close to the distal edge of the SOBP but still on the plateau. The results are plotted against both the time averaged current I_V and the effective current $I_{V,eff}$ as defined in equation (F.9) to show that there is a substantial difference and that the volume

⁸ The residual range R_{res} is the distance to the depth where the dose is reduced to 10% of the maximum dose (IAEA TRS-398).

recombination parameter would be incorrectly derived if I_V were used in equation (F.7). At this depth, the effective current is about a factor 4.5 higher than the time averaged current. At the distal edge of the Bragg peak, the effective current becomes a factor 8 higher.

Chamber	Monitor	Beam	R_{res} (cm)	A/V	$m^2 g/V^2$ (nA ⁻¹)	$g^{(1)}$ (cm)	$m^{(1)}$ (cm nA ⁻¹ V ⁻²)	$g^{(2)}$ (cm)	$m^{(2)}$ (cm nA ⁻¹ V ⁻²)
Markus SN478	Markus SN2225	Full energy	1.28	2.84E-03 (3%)	2.25E-03 (4%)	6.04E-03	3.73E+03	4.60E-03	4.90E+03
Markus SN2225	Markus SN478	"	1.48	2.44E-03 (6%)	2.48E-03 (2%)	6.04E-03	4.10E+03	4.40E-03	5.63E+03
Roos SN133	Roos SN227	"	2.08	2.28E-03 (8%)	1.94E-04 (9%)	6.63E-04	2.93E+03	6.86E-04	2.84E+03
Roos SN227	Roos SN133	"	2.66	1.76E-03 (23%)	1.85E-04 (8%)	6.63E-04	2.79E+03	6.82E-04	2.71E+03
NACP02 SN5205	NACP02 SN8508	"	1.28	2.12E-03 (21%)	4.97E-04 (20%)	1.70E-03	2.93E+03	1.34E-03	3.70E+03
NACP02 SN8508	NACP02 SN5205	"	0.96	1.12E-03 (8%)	9.44E-04 (3%)	1.70E-03	5.56E+03	1.16E-03	8.12E+03
NE2561 SN193	NACP02 SN5205	"	2.69	1.31E-03 (15%)	3.80E-04 (7%)	4.22E-03	3.60E+03	4.58E-03	3.32E+03
Markus SN478	Markus SN2225	Modulated	1.28	3.36E-03 (17%)	1.60E-03 (6%)	6.04E-03	2.65E+03	4.60E-03	3.48E+03
Markus SN478	Markus SN2225	"	0.58	3.62E-03 (6%)	1.82E-03 (12%)	6.04E-03	3.01E+03	4.60E-03	3.95E+03
Markus SN478	Markus SN2225	"	0.35	2.92E-03 (11%)	1.57E-03 (10%)	6.04E-03	2.60E+03	4.60E-03	3.42E+03
Markus SN478	Markus SN2225	"	0.07	3.61E-03 (27%)	1.45E-03 (6%)	6.04E-03	2.40E+03	4.60E-03	3.15E+03
Markus SN478	Markus SN2225	"	0.07	5.14E-03 (5%)	1.25E-03 (7%)	6.04E-03	2.07E+03	4.60E-03	2.73E+03

⁽¹⁾ Calculated using the nominal volume of ionisation chamber

⁽²⁾ Calculated using the volume of the ionisation chamber as derived from the calibration factor

Table F.1. Initial recombination A/V and volume recombination per unit dose rate $m^2 g/V^2$ derived for all the chambers in this work. Rough estimates of the standard deviations (given in percent) on the results were derived from the values of the three linear fits as shown in figures F.3a and F.3b. The derived values for the volume recombination parameter m^2 are given in the final column.

The resulting parameters A/V and $m^2 g/V^2$ from the measurements for all chambers are shown in table F.1. These are the relevant parameters for each individual chamber to calculate the recombination correction factor from the ionisation current. Provided the exact geometry of the sensitive air volume of the ionisation chamber is known, the volume recombination parameter m^2 can be derived. This is done for all the chambers as shown in table F.1. The average value for all chambers in the non-modulated beam using the nominal volumes is $m^2 = (3.7 \pm 0.4) 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$, whereas the average value using the calibration factor derived volumes is $m^2 = (4.5 \pm 0.7) 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$. Both values are, within the experimental uncertainties, in agreement with the value of $m^2 = (3.97 \pm 0.07) 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$ reported by Boutillon (1998) for X-rays. The measurements in the modulated beam (based on effective current) reveal no significant dependence on depth and yield an average value of $m^2 = (2.6 \pm 0.2) 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$ using the nominal chamber volume and $m^2 = (3.4 \pm 0.2) 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$ using the calibration factor derived volume. These values are lower than those in the non-modulated beam, but it must be noted that this was only measured for one chamber while it is clear from the measurements in the non-modulated beam that there is considerable variation from one chamber to another. It might be argued that the last row in table F.1 is not in line with the other data points given the higher initial and lower volume recombination and indeed the data as a function of current (as in figure F.3d) show a slight curvature, not present in other data sets. The values obtained by ignoring this data point are slightly, but not significantly, higher.

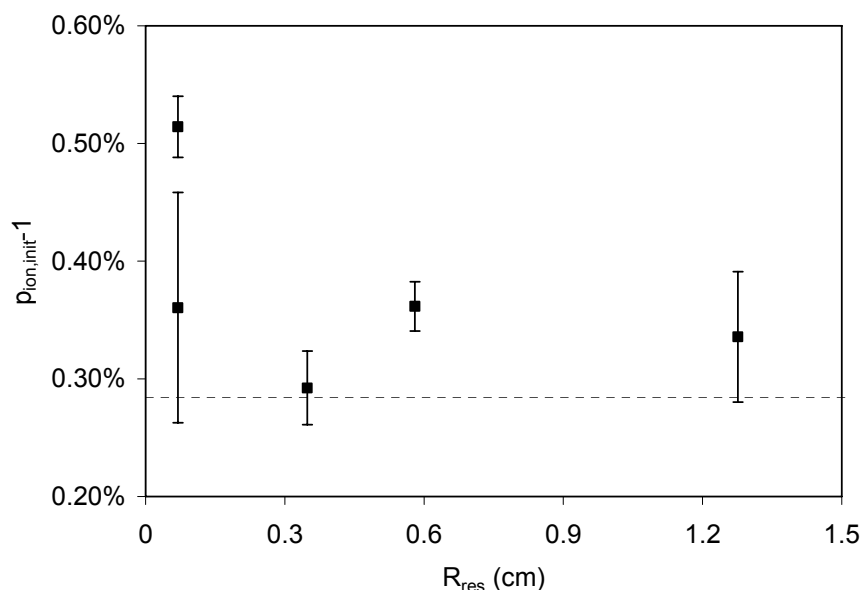


Figure F.5. Initial recombination correction for the Markus SN 478 as a function of residual range in a modulated beam. The dashed line gives the initial recombination for this chamber determined in the non-modulated beam.

In the non-modulated beam, the A/V values representing the initial recombination term are consistent per chamber type, with a slightly larger variation for the NACP02 chambers. Overall, they vary between 0.1% and 0.3%, which is comparable to the values usually found in photon and electron beams (0.1-0.2%). The results for the modulated beam are plotted in figure F.5 as a function of residual range. The result for the non-modulated beam is plotted as a dotted line and shows that there is no significant difference between initial recombination in a non-modulated beam and a modulated beam, and that there is no significant variation as a function of depth. There is only an indication of a slight increase of the initial recombination at the distal edge of the Bragg peak (the measurement was performed twice because the fluctuations tended to be a bit larger). This might be related to the higher LET, although the effect is hardly significant. Even if this effect proves to be real, it is only relevant for depth dose measurements and its magnitude lets us conclude that it poses no problem for depth dose measurements.

Table F.2 gives values of the recombination correction factors for all chambers in this study at two dose rates in each of the non-modulated and modulated proton beam. The results in the column with heading ‘Equation (F.11) specific’ are obtained from equation (F.11) using the experimental values A/V and $m^2 g/V^2$ as tabulated in table F.2. The column with heading ‘Equation (F.11) generic’ is obtained from equation (F.11) using as generic values $A = 0.25$ V and $m^2 = 3.97 \cdot 10^3 \text{ s cm}^{-1} \text{ nC}^{-1} \text{ V}^2$ and calculating g as explained in section F.2.1 with ν derived from the calibration factor using equation (F.2). It can be seen that for all the conditions met in this work, the generic values give a good agreement (within 0.1%) with the results from chamber specific data. In the last four columns, results from the two-voltage method are shown using the models from IAEA TRS-398 for pulsed and continuous beams. Only voltage ratios of 3 and 4 are considered since a ratio of 2 is not recommended in IAEA TRS-398. It is obvious that when the two-voltage method for pulsed beam would be applied, the recombination correction factor is always overestimated and the error can amount to 2%.

In general, the two-voltage method for continuous beams slightly underestimates the recombination correction, which is explained because the effect of initial recombination is not included in the model, although the resulting errors are more modest.

Chamber Beam		Equation (F.11) specific	Equation (F.11) generic	Two-voltage for pulsed beam		Two-voltage continuous beam		for
				V/3	V/4	V/3	V/4	
Full Energy Beam	$I_V^{(1)}$							
Entrance dose rate 20 Gy min ⁻¹								
Markus SN478	0.24	1.0042	1.0036	1.0105	1.0124	1.0026	1.0025	
Markus SN2225	0.25	1.0040	1.0036	1.0105	1.0127	1.0026	1.0025	
Roos SN133	1.60	1.0031	1.0036	1.0056	1.0066	1.0014	1.0013	
Roos SN227	1.61	1.0025	1.0036	1.0050	1.0056	1.0013	1.0012	
NACP02 SN5205	0.82	1.0031	1.0036	1.0061	1.0072	1.0015	1.0015	
NACP02 SN8508	0.94	1.0033	1.0036	1.0099	1.0117	1.0025	1.0024	
NE2561 SN193	1.26	1.0025	1.0027	—	—	—	—	(2)
Entrance dose rate 32 Gy min ⁻¹								
Markus SN478	0.60	1.0050	1.0043	1.0156	1.0181	1.0038	1.0036	
Markus SN2225	0.62	1.0049	1.0043	1.0162	1.0189	1.0040	1.0037	
Roos SN133	4.00	1.0035	1.0043	1.0081	1.0094	1.0020	1.0019	
Roos SN227	4.03	1.0029	1.0043	1.0069	1.0084	1.0017	1.0017	
NACP02 SN5205	2.04	1.0037	1.0043	1.0099	1.0126	1.0025	1.0025	
NACP02 SN8508	2.36	1.0047	1.0043	1.0176	1.0219	1.0043	1.0043	
NE2561 SN193	3.15	1.0032	1.0035	1.0098	1.0122	1.0024	1.0025	
Modulated Beam	$I_{V,eff}^{(1)}$							
Mid-SOPB dose rate 16 Gy min ⁻¹								
Markus SN478 at $R_{res} = 1.28$ cm	0.38	1.0050	1.0044	1.0099	1.0111	1.0025	1.0022	
" at $R_{res} = 0.58$ cm	0.62	1.0066	1.0055	—	—	—	—	(2)
" at $R_{res} = 0.35$ cm	0.81	1.0063	1.0065	1.0149	1.0172	1.0037	1.0034	
" at $R_{res} = 0.07$ cm	0.84	1.0069	1.0066	1.0162	1.0190	1.0040	1.0038	
" at $R_{res} = 0.07$ cm	0.84	1.0080	1.0066	1.0153	1.0173	1.0038	1.0034	
Mid-SOPB dose rate 26 Gy min ⁻¹								
Markus SN478 at $R_{res} = 1.28$ cm	1.01	1.0060	1.0055	1.0139	1.0161	1.0034	1.0032	
" at $R_{res} = 0.58$ cm	1.64	1.0085	1.0074	1.0222	1.0236	1.0054	1.0046	
" at $R_{res} = 0.35$ cm	2.16	1.0084	1.0090	1.0234	1.0265	1.0056	1.0051	
" at $R_{res} = 0.07$ cm	2.24	1.0089	1.0092	1.0251	1.0289	1.0060	1.0056	
" at $R_{res} = 0.07$ cm	2.24	1.0097	1.0092	1.0197	1.0236	1.0048	1.0046	

⁽¹⁾ estimated current derived from the dose rate via the ionisation chamber calibration factor - there were fluctuations from chamber to chamber in the real current at which the 2-voltage technique was applied but those have a minor effect

⁽²⁾ not measured

Table F.2. Comparison of numerical values for the recombination correction factors obtained using various methods at the two highest dose rates used in this work for both the non-modulated and the modulated proton beam and for all ionisation chambers used in this work.

The Markus is a chamber type routinely used for dosimetry at CCO. Picking out the result for the Markus SN478 chamber at a residual range of 0.6 cm, using the parameters from table F.1, the recombination correction in the modulated beam at a treatment dose rate of 26 Gy min⁻¹ (corresponding to an effective ionisation current of about 2.7 nA) is 1.0085.

F.5. Discussion and conclusions

An experiment was performed in the CCO proton beam to evaluate recombination correction factors for ionisation chambers. For plane-parallel chambers a ‘face-to-face’ set up was used to give highly correlated responses in the test chamber and monitor chamber. In this way type A uncertainties for individual data points could be kept at the level of 0.01%. This method could be applied in high-energy photon and electron beams as well. We are not aware of any previously reported results using this method.

It was demonstrated that the CCO beam should be considered as a pure continuous beam with respect to recombination. The volume recombination parameter was found to be in good agreement with the values for X-rays. Initial recombination was found to be independent of beam quality except for a slight indication of an increased initial recombination at the distal edge of the Bragg peak. This is only relevant for depth dose measurements and its magnitude permits the conclusion that it is of no concern.

For the CCO beam, where therapeutic dose rates are as high as 26 Gy min⁻¹ the recombination corrections are substantial. For a Markus chamber at the centre of the SOBP we find a correction of 0.8% consisting of a contribution of 0.3% from initial recombination and 0.5% from volume recombination. Using equation (F.11) with generic values for the initial and volume recombination parameters ($A = 0.25$ V and $m^2 = 3.97 \cdot 10^3$ s cm⁻¹ nC⁻¹ V²) the experimental results are reproduced within 0.1%.

Care must be taken with the recommendations in IAEA TRS-398 since only the formulae for pulsed beams are discussed in the section on proton beam dosimetry. Particularly, no other option is indicated in the section on protons and in the worksheet, although in the introductory sections a remark is made that proton beams from synchrocyclotrons might have to be treated as continuous beams. For the CCO beam and similar proton beams operating at high dose rates for treating optical targets, the recombination correction factor can be overestimated by up to 2% if the recommendation from IAEA TRS-398, which is only valid for pulsed beams, is followed without reflection.

F.6. References

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ADDENDUM G: PALMANS (2006A)

PALMANS, H., Perturbation factors for cylindrical ionization chambers in proton beams. Part I: Corrections for gradients. *Phys. Med. Biol.*, 2006a, **51**,3483-3501.

G.1. Abstract

An analytical model is presented to calculate the effective depth in water of cylindrical ionization chambers in clinical proton or heavy ion beams in the presence of modest and linear gradients. This model is compared with Monte Carlo simulations and recommendations in IAEA TRS-398 for all ionization chambers listed in that report. A refinement of the analytical model allows applying it for depth-dose curves even in the non-linear gradient region of the Bragg peak. Combined with information from the Monte Carlo simulations it also allows one to solve the inverse problem of deriving a depth-dose curve in homogeneous water from the depth-dose response obtained with a cylindrical ionization chamber.

The results show that the effective depth in water calculated analytically, calculated by Monte Carlo and derived from experimental literature are in good agreement. The agreement with the IAEA TRS-398 recommendation is good for most ionization chambers but can be wrong by up to 1.5 mm for some ionization chamber types. The agreement between the refined analytical model and Monte Carlo simulations of depth-dose response curves is shown to be good (even in extreme situations of large ionization chambers in low-energy beams). An indication is given of situations in which a simple shift of the entire depth-dose curve is sufficiently accurate. It is demonstrated that it is possible to solve the inverse problem with this method, even for rather noisy data. This is illustrated by successfully applying it to depth dose measurements with cylindrical ionization chambers taken from the literature.

G.2. Introduction

The increasing use of proton and ion beams for radiotherapy, and the consequential need for improved reference dosimetry, has led to the publication of a series of recommendations for dosimetry of these beams using an ionization chamber. AAPM report 16 (Lyman et al. 1986), the ECHED code of practice and its supplement (Vynckier et al. 1991, 1994), ICRU report 59 (ICRU 1998) and IAEA TRS-398 (Andreo et al. 2000) all recommend cylindrical ionization chambers as reference instruments. IAEA TRS-398 also provides data for the use of plane-parallel ionization chambers for low-energy proton beams, even though plane-parallel ionization chambers have not as yet been extensively characterised for protons. Some recent studies on the use of plane-parallel ionization chambers as reference dosimeters have been published (Vatnitsky et al. 2002, 2003, Palmans et al. 2002, Kacperek et al. 2003, Kanai et al. 2004), but since the uncertainties on perturbation correction factors for plane-parallel ionization chambers in the calibration beams are usually large, the preferred option remains to use cylindrical ionization chambers.

In all the codes of practice quoted above, the centre of the cylindrical ionization chamber is taken as reference point for dosimetry. In general it is recommended to perform reference dosimetry in the absence of a depth-dose gradient, i.e. in the entrance region for non-modulated beams or in the middle of the spread-out Bragg peak (SOBP), such that volume averaging effects do not play a role, however, this is not always possible. In the entrance

region of a non-modulated Bragg peak for example, there can be a slight gradient, which can be even negative at higher proton energies. In potential future functional treatments where the (inhomogeneous) biological response of tissues could be taken into account, the SOBP will not necessarily be flat either. In ion beams, it is even standard practice to take into account that the biological response depends strongly on the particle energy by deliberately introducing a gradient to the SOBP.

For depth-dose curve measurements in the Bragg peak region, one will also have to deal with volume averaging effects. It can be argued that plane-parallel ionization chambers are advised for depth dose measurements, but in the presence of strong gradients such as in a low-energy Bragg peak, the unavoidable deviation from perpendicularity to the beam axis, will to some extent also induce volume averaging. Potential depth dependent polarity effects of some plane-parallel ionization chamber (as reported in electron beams, see e.g. IAEA TRS-398) provide another reason why small cylindrical ionization chambers could be preferred above plane-parallel ionization chambers for depth dose measurements. Palmans (2000) showed that an analytical model gave a reasonable agreement with Monte Carlo simulations of the depth-dose response of an FWT IC-18 ionization chamber. But the statistics were not very good due to a lack of available CPU time for the Monte Carlo simulations to simulate this problem even for one ionization chamber.

Effects due to the presence of depth-dose gradients can be accounted for in two ways. A gradient (or displacement) perturbation correction factor can be applied or alternatively the concept of effective point of measurement, P_{eff} , can be used. P_{eff} is defined as the point in homogeneous water where the particle fluence is the same as in the air cavity. In theory and generally speaking there is no such point, but it could practically be defined as the point in homogeneous water where the particle fluence will result in the same dose to air as the particle fluence in the air cavity. Strictly speaking, the definition of P_{eff} only applies to a theoretical air cavity surrounded by water, whereas in practice the central electrode, wall and waterproofing sleeves also induce gradient effects (apart from secondary particle perturbations which are accounted for in p_{cel} and p_{wall} correction factors). To denote the total gradient related effect of cavity, central electrode, wall and sleeve, we use the term of effective depth in water, $z_{\text{w,tot}}$, of the ionization chamber. As was shown by Palmans and Verhaegen (2000) the meaning of P_{eff} and $z_{\text{w,tot}}$ is often confused and the distinction was made clear above in order to avoid this confusion in the remainder of this paper.

The gradient correction factor can in principle be obtained experimentally as the ratio of the reading with the centre of the ionization chamber at the reference depth and the reading with $z_{\text{w,tot}}$ at the reference depth, provided the latter is known. Some authors have measured the effective depth in water by comparing depth-dose curves obtained with cylindrical ionization chambers to those obtained using plane-parallel ionization chambers (Jäkel et al. 2000, Mobit et al. 2000 and Kanai et al. 2004). Note that this list of experiments contains results for both proton and carbon beams. We assume that heavy ions can be treated in the same way as protons with respect to gradients. Hartmann et al. 1999 used a different method; they determined the shift that had to be applied to a cylindrical ionization chamber measurement such that the calibration of the monitor was consistent as a function of beam energy. In all four references, the experimental shift of $z_{\text{w,tot}}$ with respect to the centre of the ionization chamber, $\Delta z_{\text{w,tot}}$, was close to the value of 0.75 times the radius of the cylindrical cavity, which is recommended in IAEA TRS-398 for heavy ion beams. However, Palmans and Verhaegen (2000) warned that this is the case since only Farmer type ionization chambers were used, but that deviations from the IAEA value can be considerable for other ionization chamber types.

Palmans and Verhaegen (1998), Jäkel et al. (2000) and Kanai et al. (2004) used an analytical integration over the cavity geometry to obtain the shift of $z_{w,tot}$ from the centre of the ionization chamber related to the cavity and wall effect. Bichsel (1995) used an analytical model to calculate the convolution of the depth-dose curve with cylindrical and plane-parallel detectors with finite sizes (such as solid state detectors). Ulmer and Kaissl (2003) solved the inverse problem for plane-parallel detectors with finite sizes using a Gaussian convolution model.

In this work, the analytical model from Palmans and Verhaegen (1998) and Palmans (2000) is extended to take the effect of the sleeve and central electrodes into account. The emphasis by Palmans and Verhaegen (1998) was in fact on the calculation of perturbation correction factors for gradient effects, but the approach adopted here is to calculate the effective depth in water of the ionization chamber, a quantity that does not depend on the magnitude of the local gradient. The results will be compared with results from Monte Carlo simulations and with the earlier quoted experimental data. The first aim is to apply the analytical model for the calculation of $\Delta z_{w,tot}$ for all commercial cylindrical ionization chambers listed in IAEA TRS-398. In addition, it will be demonstrated that a slight modification to the model allows correcting depth-dose curves measured with a cylindrical ionization chamber even in the strongly non-linear gradient region of the Bragg peak, if the size of the ionization chamber is restricted to certain limits. The results of this model will be compared with the results of Bichsel (1995) and with Monte Carlo simulated results. It will be shown that the analytical model can be applied to the inverse problem as well, i.e. the reconstruction of the depth-dose curve from the depth-dose function measured with a cylindrical ionization chamber.

G.3. Method

G.3.1. Analytical model for the effective depth in water of an ionization chamber

Palmans and Verhaegen (1998) showed an analytical model for the effective depth of a cylindrical or spherical cavity surrounded by a wall submersed in water, which is extended here. Four assumptions are made:

- (i) protons travel along straight lines,
- (ii) the distance a proton travels through a medium can be converted to a water equivalent distance by multiplying with the ratio of the $csda^9$ ranges in water and the medium (see e.g. Palmans and Verhaegen 1997),
- (iii) processes like multiple scattering, energy straggling and nuclear absorption are not substantially altered by the presence of non-water materials over the distances relevant in this model (e.g. the wall thickness) such that the contributions from these processes to the exact shape of the depth-dose curve are inherently taken into account,
- (iv) the depth-dose gradient is approximately linear on the local scale of the ionization chamber dimensions.

Under these assumptions an unambiguous value of the effective depth can be calculated by integration over the projected area of the cavity perpendicular to the beam direction. A brief outline of this analytical model is given in the appendix (section G.6.1) and the development and more details are described by Palmans (2006).

In the presence of a substantial non-linearity of the gradient, such as in the vicinity of the Bragg peak, assumption (iv) in the previous section is not fulfilled and the integration needs

⁹ Continuous slowing down approximation

to be kept explicitly in the equations resulting in an extended analytical model. In this case it is no longer possible to assign a single and unambiguous effective depth to account for the gradient effect. A brief outline of this extension is presented in the appendix (section G.6.2) and a more detailed discussion can again be found in Palmans (2006).

G.3.2. Monte Carlo simulations

Monte Carlo simulations of the gradient effects were performed with McPTRAN.MEDIA and McPTRAN.CAVITY, which have both been described by Palmans (2004). These Monte Carlo codes use the transport algorithm of PTRAN (Berger, 1993) that simulates proton pencil beams in homogeneous water. McPTRAN.MEDIA allows transport in materials other than water and also in-line calculation of stopping power ratios. The latter feature has been used for calculating dose to air in homogeneous water geometry by calculating the air to water stopping power ratio as a function of depth. McPTRAN.CAVITY allows the simulation of more complex geometries as shown in figure G.1. The most important aspect is the transport through a series of embedded cavities that can have spherical, cylindrical or thimble geometries. As shown in figure G.1, the modelling of a range modulator wheel is implemented as well, but this feature is not used in this work. A geometry interrogation region is introduced such that the tests to check if a particle can enter or leave a cavity do not have to be performed throughout the entire simulation. The energy of the incident proton can be sampled from a Gaussian distribution, but all protons are set in motion on the same axis.

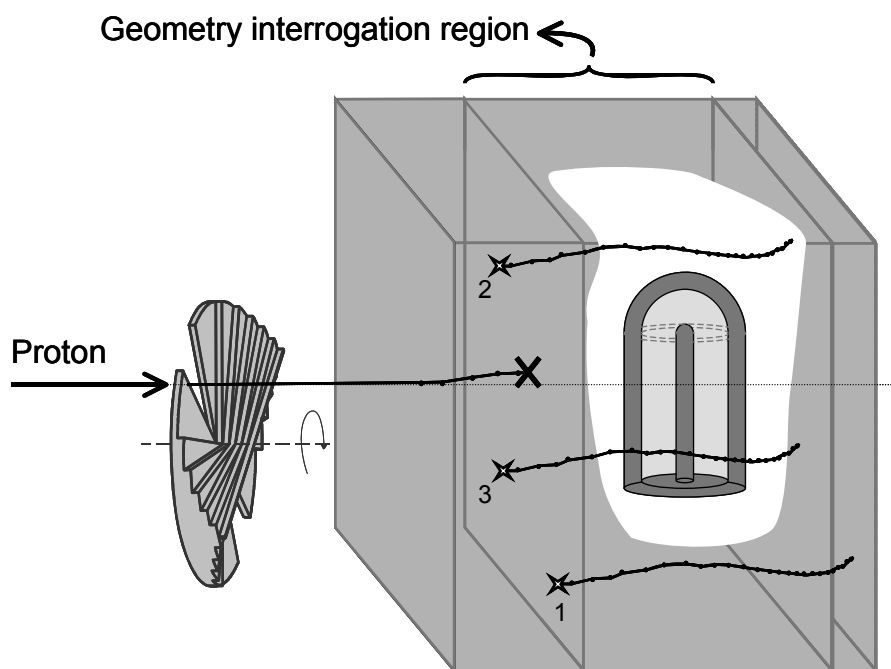


Figure G.1. Geometry used for the Monte Carlo simulations.

Apart from the geometry interrogation method, other variance reduction techniques have been used of which the most important are “history splitting” and “lateral range rejection”. The former is implemented as follows: upon reaching the geometry interrogation region (as marked by a black cross in figure G.1), all transport parameters of the particle are stored and the transport is resumed a number of times with lateral displacements that are sampled from the field geometry of the simulated beam, which can be rectangular or circular (as an

example, three of the resumed histories are shown in figure G.1 indicated with labels 1, 2, and 3). This way the field size is taken into account. The repetition number is calculated such that on the average, the outer cavity will be hit once and the weight of every proton is reduced according to the number of history repetitions. For the second variance reduction method, the maximum lateral displacement of the particle over the thickness of the geometry interrogation region is calculated upon resuming each of the histories. If this happens to be smaller than the lateral distance to the outer cavity, the history is not simulated any further. In addition, the whole procedure started at the beginning of the geometry interrogation region can be repeated for an arbitrary number of cavity sets representing different ionization chambers or detectors. These variance reduction techniques result in a gain of a factor of 20 to 100 in computing time depending on the proton energy. After the transport for all the ionization chambers at one depth is finished, the original history (prior to entering the geometry interrogation region) is resumed with the cavity and interrogation region positioned at a new depth. This way the depth-dose response can be calculated for the ionization chambers in one simulation. Typically 50 or 120 depths are specified in one simulation with a higher spatial resolution in the Bragg peak than in the entrance region.

Simulations were performed for all 52 cylindrical ionization chamber types tabulated in IAEA TRS-398 for proton energies of 60 MeV, 80 MeV, 100 MeV, 150 MeV and 200 MeV. All ionization chambers were modelled as thimble shaped. This is an approximation for some ionization chambers that have a conical instead of a hemispherical top end, but this usually concerns ionization chambers like Farmer types where the cylindrical part is the dominant contribution and so we assume that errors due to this approximation are small. The length of the cavity (cylinder + hemisphere), the cavity radius and the wall thickness were taken from IAEA TRS-398, but the central electrode dimensions are not listed in that report. Since most central electrodes have a diameter close to 1.0 mm, this was taken as a general figure, except for those ionization chambers where we have information from the manufacturer that it deviates significantly from 1.0 mm. Hence, the central electrode diameters were taken to be 4.6 mm for the Exradin A2 and T2, 2.0 mm for the FWT IC-18, 3.0 mm for the NE2581, 1.5 mm for the PTW23331 and 2.0 mm for the PTW23332. For the NE2561/NE2611, which have a hollow central electrode of about 1.8 mm external diameter, it was assumed that this is the equivalent of a 1.0 mm diameter solid electrode. Furthermore, for ionization chambers where the manufacturers data show substantial deviations from the dimensions given in IAEA TRS-398 it was verified whether or not these differences have a significant influence on the results. This includes the NE2561/NE2611 ionization chamber, but also a number of PMMA walled ionization chambers that have a conducting graphite layer or graphite/epoxy paste on the inner surface of the cavity. For all non-waterproof ionization chambers, a 0.5 mm PMMA sleeve was assumed and included in the simulations as was done for generating the data in IAEA TRS-398.

The simulations were run on a distributed computing grid, which consists of the 200 fastest desktop PCs at NPL, which accept jobs from a server for the time they are not occupied by a desktop user application. The Fortran code was compiled with g77 under Cygwin, a Linux emulator under Windows. A typical simulation for one proton energy and all 52 ionization chamber types at 50 depths took 1.5 days on the grid consuming about 5600 hours of CPU time (~equivalent on a P4 2GHz). Such an extensive set of simulations would not have been possible without availability of the distributed computing grid.

G.4. Results and discussion

G.4.1. Comparison of the analytical model results with Monte Carlo calculations

G.4.1.1. Effective water depth in entrance and distal regions

Figure G.2 shows for some ionization chamber types and some energies the dose in air, D_{air} , as a function of depth in water, the same curve shifted in depth by the analytical $\Delta z_{\text{w,tot}}$ and D_{air} in the ionization chamber's air cavity from the Monte Carlo simulation. All calculations except one were done for mono-energetic mono-directional proton beams. The results for mono-energetic beams are worst cases (or best cases to test the model) since clinical beams will at least have a certain energy and angular spread and will commonly be spread-out as well leading to smoother Bragg curves with smaller peak to entrance dose ratios. Comparing figures G.2c and G.2d shows for example that a Gaussian energy spread with a standard deviation of 1% of the energy makes already a large difference. The type-A (statistical) uncertainty on the Monte Carlo results was smaller than 0.05% on all data points except those in the tail of the distal edge of the Bragg peak. It is obvious that up to a certain depth a single shift, $\Delta z_{\text{w,tot}}$, is adequate. The example for the Exradin T1 shows that using a small ionization chamber it can be even sufficient at any depth (except that it leads to a small underestimation of the measured dose maximum). A criterion for this can be derived from the simulations of all ionization chambers: for those with a ratio of central electrode diameter and cavity diameter smaller than 0.25 and a radius smaller than 4 mm the differences will be smaller than 3% of the maximum dose above 150 MeV (the difference being mainly in the Bragg peak). At low energies, on the other hand, the Bragg peak in the Monte Carlo results is heavily distorted for all ionization chambers due to volume averaging in the ionization chamber geometry and thus a simple shift is too crude in that region. Figure G.2 gives an impression of the range of depths where the more detailed model of section G.6.2 will be required.

Values for $\Delta z_{\text{w,tot}}$ can be derived from the Monte Carlo simulations by calculating the distance from the centre of the ionization chamber to the depth where D_{air} in homogeneous water (in a point) equals the average D_{air} in the air cavity. This can of course not be done in the distorted depth dependence of dose to air in the Bragg peak region, but it is equally difficult to get an accurate value if the gradient is very small such as at shallow depths in the entrance region. For these reasons, the following rather arbitrary criterion was used: only depths where the local gradient of the ionization chamber depth dose is between 3% per mm and 10% per mm were considered. This interval has been indicated in the plots of figure G.2. Only for one ionization chamber, the Exradin A2, an exception was made; since the distortion of the Bragg peak by this ionization chamber extends to a very wide range of depths, overlapping considerably with the range defined by the criterion, the upper limit was set to 8% per mm. The results are averaged over all calculated points that pass the criterion. Since there was no obvious trend of the result as a function of energy, an average over the five energies was taken. The standard deviation of the results was smaller than 0.03% for all ionization chambers.

Another possibility is to derive $\Delta z_{\text{w,tot}}$ from the shift of the distal edge of the Bragg peak. This requires a criterion for the dose level at which this is evaluated. Since usually the depth where the dose has dropped to 80% of the dose maximum, z_{80} , is used as a reference to determine the csda range we have chosen to evaluate the shift at this dose level. In this case there was an obvious dependence on energy with the values being larger for lower energies and reaching an asymptotic value at higher energies. Only values above this asymptotic value, which depends on the ionization chamber type, were taken into account.

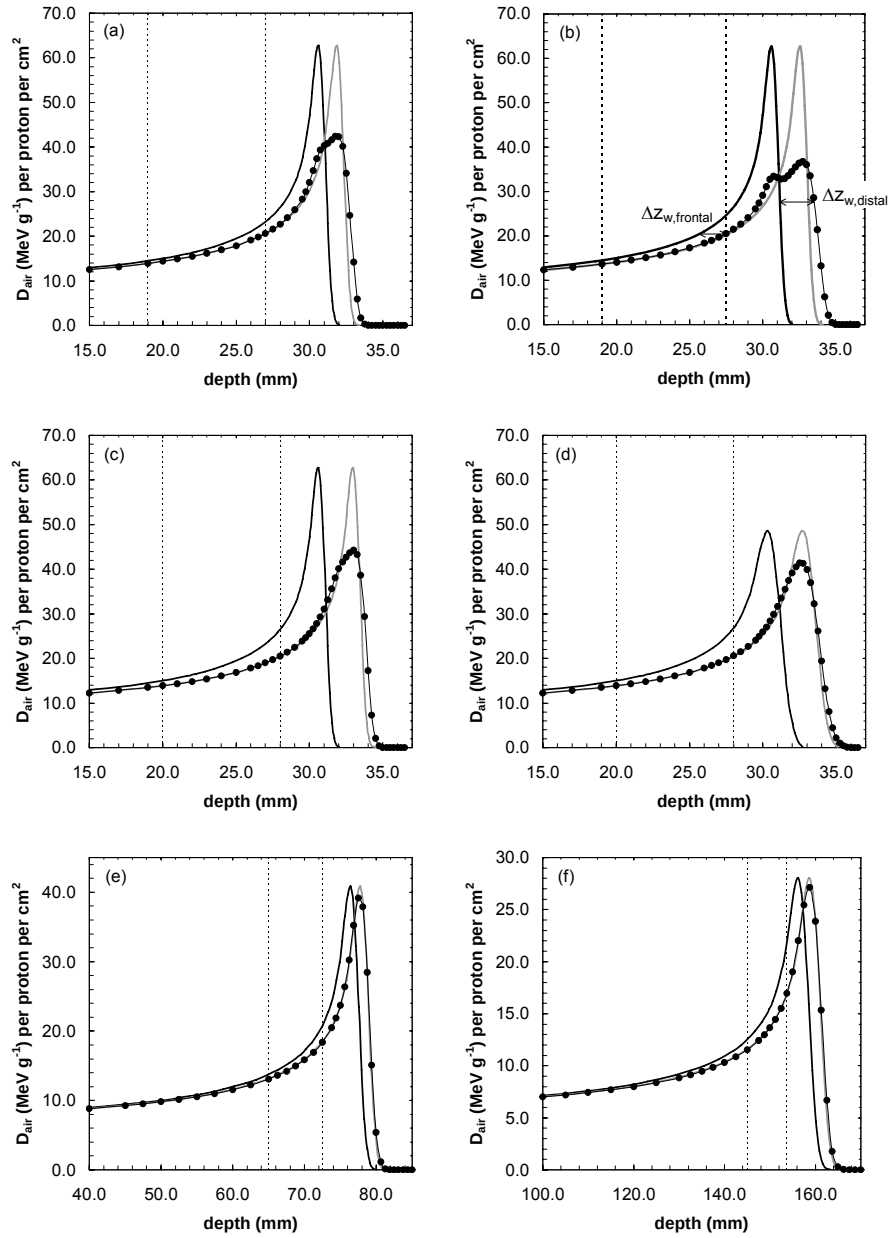


Figure G.2. Depth dose to air curve calculated by Monte Carlo in homogeneous water (black curve), depth dose to air curve shifted over $\Delta z_{w,tot}$ (grey curve) and depth dose to air as a function of depth for a thimble ionization geometry (symbols connected by thin lines), (a) for an FWT-IC18 in a 60 MeV mono-energetic beam, (b) for an NE2581 in a 60 MeV mono-energetic beam, (c) for a PTW-30006 in a 60 MeV mono-energetic beam, (d) for a PTW-30006 in a 60 MeV beam with a Gaussian energy spread ($1\sigma = 1\%$), (e) for an Exradin T1 in a 100 MeV mono-energetic beam and (f) for an NE2561/NE2611 in a 150 MeV mono-energetic beam. The vertical dashed lines indicate the interval where the local gradient is between 3% and 10% of the local dose per mm.

Type of IC	-Δz _{w,tot} (mm)				Type of IC	-Δz _{w,tot} (mm)			
	This work			IAEA		This work			IAEA
	MC _f	MC _d	Analyt.			MC _f	MC _d	Analyt.	
Capintec PR05P mini	0.9	0.7	0.7	1.5	PTW23323	0.8	0.8	0.8	1.2
Capintec PR05 mini	0.8	0.7	0.7	1.5	PTW23331	3.0	3.0	3.0	3.0
Capintec PR06C	2.4	2.4	2.4	2.4	PTW23332	1.3	1.5	1.4	1.9
Exradin A2	2.0	2.1	2.1	3.6	PTW23333	2.4	2.3	2.4	2.3
Exradin T2	2.8	2.9	2.8	3.6	PTW30001	2.4	2.4	2.4	2.3
Exradin A1	1.0	0.9	0.9	1.5	PTW30002/11	2.2	2.1	2.1	2.3
Exradin T1	1.5	1.4	1.4	1.5	PTW30004/12	2.1	2.1	2.1	2.3
Exradin A12	2.2	2.2	2.2	2.3	PTW30006/10/13	2.4	2.3	2.4	2.3
FWT IC18	1.3	1.2	1.2	1.7	PTW31002	2.2	2.0	2.0	2.1
FZH TK01	2.6	2.5	2.6	2.6	PTW31003	2.1	2.1	2.1	2.1
NA30-750	1.4	1.2	1.2	1.5	SNC100730	2.7	2.7	2.7	2.6
NA30-749	2.3	2.0	2.0	2.3	SNC100740	2.4	2.4	2.4	2.6
NA30-744	2.3	2.1	2.1	2.3	Victoreen RadoconII 555	1.8	1.8	1.8	1.8
NA30-716	2.2	2.1	2.1	2.3	Victoreen RadoconIII 550	0.7	0.4	0.4	1.9
NA30-753	2.3	2.2	2.2	2.3	Victoreen 30-348	1.8	1.8	1.8	1.9
NA30-751	2.3	2.3	2.3	2.3	Victoreen 30-351	2.4	2.3	2.4	2.3
NA30-752	2.2	2.2	2.1	2.3	Victoreen 30-349	3.1	3.1	3.1	3.0
NE2515	2.3	2.1	2.1	2.3	Victoreen 30-361	1.6	1.6	1.6	1.8
NE2515/3	2.4	2.2	2.2	2.4	Wellhöfer IC05	2.3	2.0	2.0	2.3
NE2577	2.3	2.2	2.2	2.4	Wellhöfer IC06/CC08	2.3	2.0	2.0	2.3
NE2505	2.2	2.2	2.2	2.3	Wellhöfer IC10	2.3	2.1	2.1	2.3
NE2505/A	2.3	2.2	2.3	2.3	Wellhöfer IC15/CC13	2.3	2.1	2.1	2.3
NE2505/3	2.3	2.3	2.3	2.4	Wellhöfer IC25/CC25	2.2	2.1	2.1	2.3
NE2571	2.3	2.3	2.3	2.4	Wellhöfer IC28/FC23-C	2.3	2.2	2.2	2.3
NE2581	1.9	2.0	2.0	2.4	Wellhöfer IC69/FC65-P	2.3	2.3	2.3	2.3
NE2561/NE2611	2.6	2.5	2.5	2.8	Wellhöfer IC70/FC65-G	2.2	2.2	2.2	2.3

Table G.1. $-\Delta z_{w,tot}$ in mm for all ionization chambers (ICs) listed in IAEA TRS-398, according the analytical model, the Monte Carlo simulations (frontal: MC_f and distal: MC_d) and the general recommendation of 0.75 times the cavity radius given in IAEA TRS-398. The ionization chambers for which the difference with the IAEA model is more than 0.5 mm are highlighted in italic, those for which the difference is more than 1 mm in bold.

Table G.1 shows the values of $(-) \Delta z_{w,tot}$ obtained from the analytical model and from the Monte Carlo simulations using both methods explained in the two previous paragraphs. The results are compared with the general recommendation in IAEA TRS-398 to use 0.75 times the radius of the cavity. Both the frontal and distal results are in general in good agreement with the analytical results. For those ionization chambers of which the cylindrical section constitutes the largest part of the volume and the central electrode has a small diameter compared to the cavity diameter, the agreement is excellent. For other ionization chambers it

can be worse, in particular when the cylindrical section is not much longer or even shorter than the diameter of the cavity as well as when the central electrode diameter is large compared to the cavity dimensions. The deviation from the IAEA recommendation is small for most ionization chambers but can be considerable (up to 1.5 mm) for some chambers commonly used in proton dosimetry such as the FWT IC-18. For Farmer type ionization chambers as well as for most ionization chambers with a thin wall and a central electrode diameter that is small compared to the cavity diameter, the agreement is good in general, consistent with the observations of Jäkel et al. (2000) and Palmans and Verhaegen (2000).

Type of IC	R_{cav} (mm)	R_{cel} (mm)	Wall	$-\Delta z_{\text{w,tot}}$ (mm)			
				This work			IAEA
				MC _f	MC _d	Analyt.	
NE2561/2611	3.675	0.875 ^(*)	0.5 mm graphite	2.4	2.5	2.5	2.8
PTW30006/10/13	3.05	0.55	0.09 mm graphite + 0.34 mm PMMA	2.2	2.2	2.3	2.3
PTW30001	3.05	0.50	0.15 mm graphite/epoxy-paste + 0.28 mm PMMA	2.4	2.4	2.2	2.3
PTW23331	3.95	0.75	0.15 mm graphite/epoxy-paste + 0.40 mm PMMA	2.9	3.0	2.9	3.0
PTW23332	2.50	1.00	0.15 mm graphite/epoxy-paste + 0.35 mm PMMA	1.3	1.6	1.3	1.9

^(*)Hollow with inner radius of 0.7mm

Table G.2. $-\Delta z_{\text{w,tot}}$ in mm for some ionization chambers that have characteristics different from the ones specified in IAEA TRS-398.

Table G.2 shows similar results for some ionization chambers of which the geometry is different from what is specified in IAEA TRS-398 (the results in the IAEA column were also based on the slightly different values of the cavity radius, to indicate the difference this gives compared to table G.1). In particular, the hollow central electrode of the NE2561/NE2611 ionization chamber and the mixed wall of a number of PTW ionization chambers could potentially result in different results. However, comparing the results for these ionization chambers in tables G.1 and G.2 it is clear that the differences are minor and that the data as specified in IAEA TRS-398 are sufficiently accurate for calculating $\Delta z_{\text{w,tot}}$.

G.4.1.2. Depth-dose distributions

Figure G.3 shows depth distributions of D_{air} in an air cavity calculated with McPTRAN.CAVITY and distributions by integrating the depth dose in homogeneous water according to the model of section G.6.2. The integration was done by dividing the central electrode sections (a) and (b) (see figure G.1) into three sub-intervals and section (c) into four sub-intervals, which is still a quite crude approach. As can be seen the model from section G.6.2 gives excellent agreement even for an ionization chamber with a thick electrode such as the NE2581 in a low-energy 60 MeV beam. The only exception is the

Exradin A2 with a thick wall and a large electrode diameter, for which more subintervals are needed for obtaining a similar agreement.

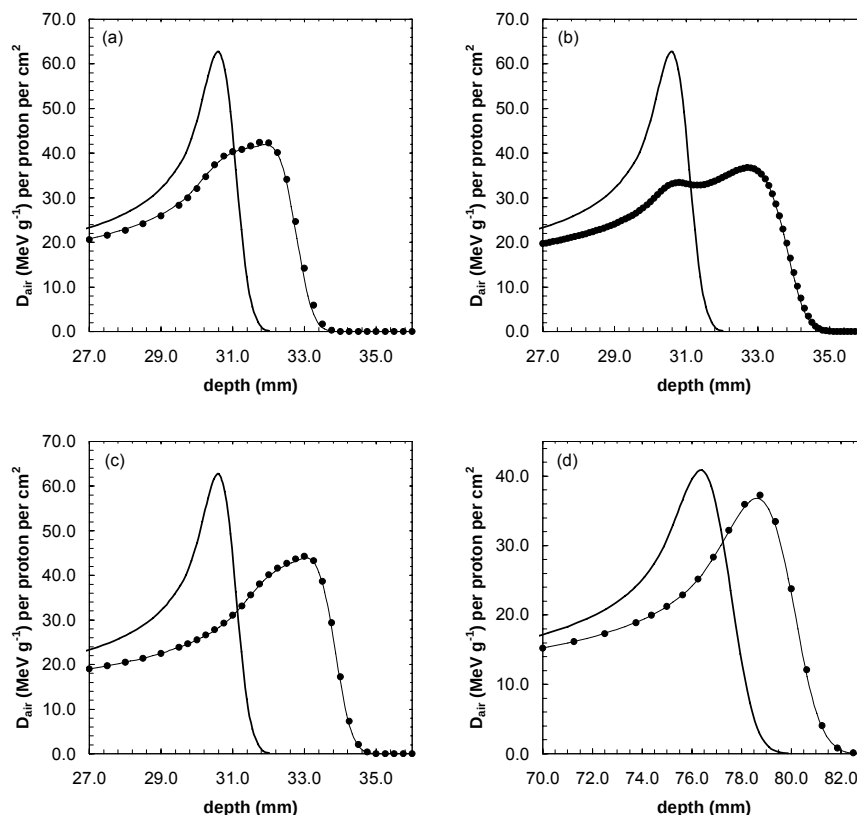


Figure G.3. Depth dose to air curve calculated by Monte Carlo in homogeneous water (black curve), depth dose to air as a function of depth for a thimble ionization geometry (symbols) and predicted depth dose to air curve calculated with the analytical model of section G.6.2 (the black curve which almost coincides with the symbols in all cases), (a) for an FWT-IC18 in a 60 MeV mono-energetic beam, (b) for an NE2581 in a 60 MeV mono-energetic beam, (c) for a PTW-30006 in a 60 MeV mono-energetic beam and (d) for a PTW-30006 in a 100 MeV mono-energetic beam.

Although an arbitrary choice such as equal volume subintervals (per section (a), (b) and (c)) gives a good agreement, the choice of the intervals was optimised in order to give the best (least squares) fit to the Monte Carlo simulations for assessing the inverse problem in the next section.

G.4.1.3. Inverse problem of estimating the depth-dose curve from the measured depth-dose response using a thimble ionization chamber

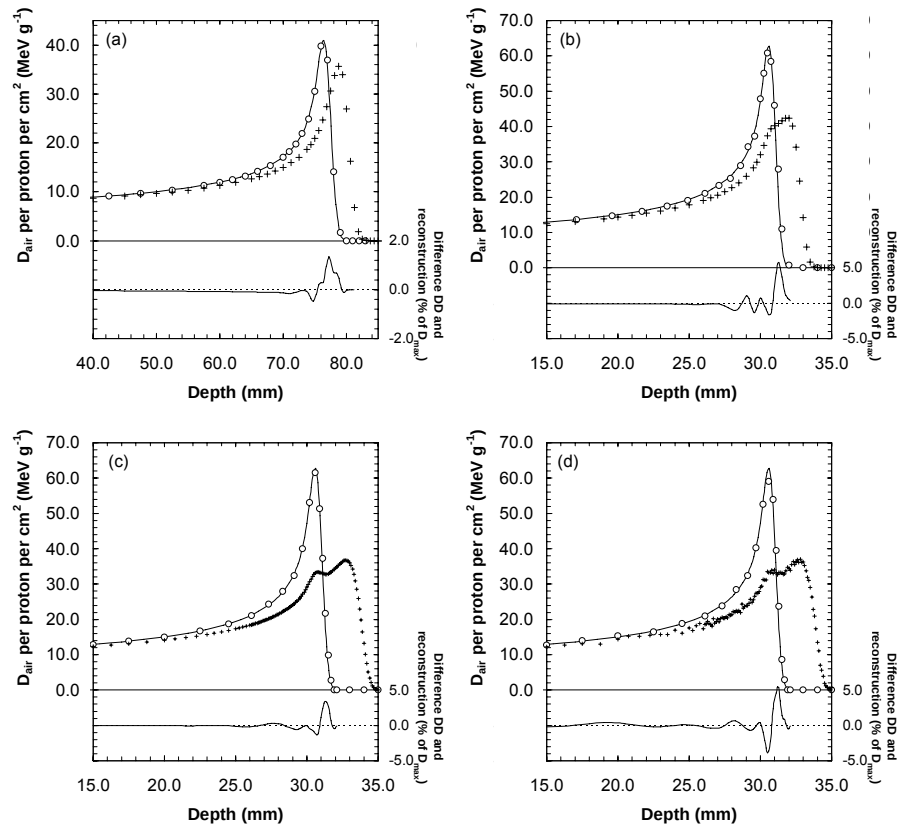


Figure G.4. Reconstruction of depth-dose curves from an ionization chamber response. The + symbols are the result from a Monte Carlo simulation of the ionization chamber response, the full curve is the Monte Carlo calculated depth-dose curve in homogeneous water and the circles are the points left as free parameters for the reconstructed depth-dose curve, (a) for an NE2561/NE2611 in a 100 MeV mono-energetic beam, (b) for an FWT-IC18 in a 60 MeV mono-energetic beam, (c) for an NE2581 in a 60 MeV mono-energetic beam and (d) for an NE2581 in a 60 MeV mono-energetic beam with deliberately added noise on the Monte Carlo simulations ($1\sigma = 2\%$). The lower panel gives the difference, expressed as a percentage of the dose maximum, between the cubic spline through the reconstructed data points and the curve in homogeneous water.

The possibility of applying the analytical model to the more challenging problem of converting a depth-dose curve obtained with an ionization chamber to the depth dose in homogeneous water is demonstrated in figure G.4 for a few ionization chambers and energies. To this end a function consisting of an arbitrary set of depth-dose values connected by a cubic spline was constructed. This function was convolved with the analytical model and a least square optimisation of the set of depth-dose values was performed in order to minimise the difference with the Monte Carlo simulated depth-dose function for the ionization chamber. In this work, this optimisation was performed with Microsoft Excel's solver. Figure G.4 shows for some chambers the resulting optimised set of data points and

the difference of the cubic spline with the depth-dose curve in homogeneous water. For most ionization chambers this gives good results with differences to the real depth-dose curve smaller than 1% of the dose maximum in the region frontal to the Bragg peak and 5% in the distal edge, where these differences are less important since they correspond to minor shifts in depth. Even when deliberately adding Gaussian noise to the Monte Carlo data points (we could as well have simulated less histories) the outcome of the inverse optimisation is still reasonable. This indicates that the method will be applicable to measured depth-dose curves, which will always exhibit a certain level of fluctuations, as well. It is again for ionization chambers with very thick, non-water equivalent walls or large electrodes that the technique is difficult to apply with satisfying results.

G.4.2. Comparison with theoretical and experimental data from the literature

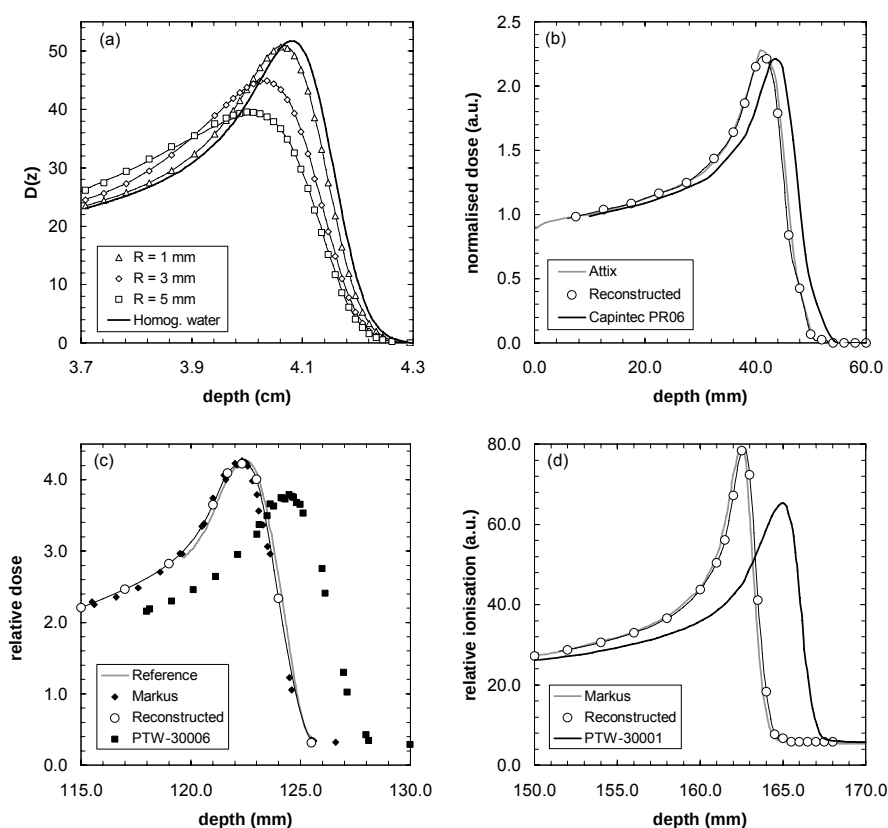


Figure G.5. (a) Comparison with the analytical model of Bichsel (1995); the lines are the results from Bichsel, the symbols the results according to the model of section G.6.2. (b) Comparison of the inverse dose reconstruction method with experimental results using a cylindrical and a plane-parallel chamber by Mobit et al. (2000), (c) idem with results by Jäkel et al. (2000) and (d) idem with results by Kanai et al. (2004).

Bichsel (1995) developed an analytical model for the simpler geometry of a pure cylindrical air cavity in homogeneous water and modelled the beam angle and energy spread widening analytically as well. The analytical model of section G.6.2. can be applied for a cavity without wall and central electrode to compare with Bichsel's results using his calculated

depth-dose curve to calculate the integrals. The cavity was divided in five strips with equidistant angles for this purpose. The results are shown in figure G.5a and as can be seen the agreement is good even though the analytical model used here is conceptually simpler and cruder.

Reference	Type of IC	Beam	$-\Delta z_{w,tot}$ (mm)			
			Experiment	This work		IAEA
				MC	Analytical	
(Mobit et al. 2000)	Capintec-PR06C	^1H	2.0 ± 0.4	2.4	2.4	2.4
(Hartmann et al. 1999)	PTW-30001	^{12}C	3 ± 1	2.4	2.2	2.3
(Jäkel et al. 2000)	PTW-30006	^{12}C	2.2 ± 0.2	2.2	2.3	2.3
(Kanai et al. 2004)	PTW-30001	^{12}C	2.5 ± 0.1	2.4	2.2	2.3

Table G.3. Experimental data from the literature on the shift (towards the beam source) of the effective depth of some cylindrical ionization chambers in mm.

As discussed in the introduction, only a few experimental determinations of the effective depth in water are available from the literature. The results of these experiments are given in table G.3 and compared with the values calculated in this work. The results are within the experimental uncertainties in agreement with the theoretical values from our work. For the PTW-30001 and PTW-30006 ionization chambers the values from table G.2, incorporating the effect of the mixed wall composition, were taken except for the IAEA values that are just the values which would be derived from the IAEA data and recommendation (listed in table G.1).

Figures G.5b, G.5c and G.5d apply the method of section G.6.2. to the experimental depth-dose curves from Mobit et al. (2000) for a 78 MeV proton beam, Jäkel et al. (2000) for a 250 MeV $\text{u}^{-1} \text{ }^{12}\text{C}$ beam and Kanai et al. (2004) in a 290 MeV $\text{u}^{-1} \text{ }^{12}\text{C}$ beam. To this end, the experimental data from these papers were digitised. For the data of Mobit et al. (2000) a small dose re-normalisation of the optimised curve of 1.6% needed to be done similar to the normalisation they did when shifting the curve. This is consistent with what one expects since, given that there must be a shift of the effective depth of the cylindrical ionization chamber, one makes an error by normalising this to the curve for the plane-parallel ionization chamber. For the data of Jäkel et al. (2000) a more substantial re-normalisation of 6% needed to be done. This is probably similar to what they have done, but no information is provided on the normalisations that were used. Even though the number of measured data points by Mobit et al and Jäkel et al. was rather limited (consequently the number of free data points that could be allowed in the optimisation was limited as well), the method still works well and given the fluctuations on the experimental data no significant differences can be observed between the reconstructed curves and the measurements using plane-parallel ionization chambers. For the data of Kanai et al. (2004), which were presented by them as not-normalised ionization curves, no additional normalisation was required. This is an indication that the assumption made in IAEA TRS-398 that the ionization chamber perturbation correction factors are unity is valid. Again the method works well, but the fine resolution of their measurements allows resolving the small shift of 0.2 mm between the reconstructed curve and the plane-parallel ionization chamber measurement. Such small

differences are of course within common levels of positioning uncertainty, but are also influenced by small differences in the stopping power ratios for protons and heavier ions.

Some additional experimental evidence was obtained by Palmans et al. (2001) who performed an experiment in which graphite build-up caps were used to increase the wall thickness of a NE2571 ionization chamber. The agreement of their experimental results with the theoretical calculations was good. In the same paper, it was shown that the application of the gradient correction factors to a number of ionization chambers brings the mutual differences between ionization chambers in better agreement with the mutual differences on the SOBP at the same depth (where it is assumed that no gradient corrections are needed). The remainder of the mutual differences was then attributed to secondary electron perturbations calculated by Verhaegen and Palmans (2001). The models used by Palmans et al. (2001) were simpler but consistent with the models in this paper.

G.5. Conclusions

An analytical model was presented to calculate the effective depth in water, $z_{w,tot}$, of cylindrical (thimble) ionization chambers in clinical proton or a heavy ion beams. The shift of this quantity from the centre of the cavity was then calculated according to this model as well as with Monte Carlo simulations for all commercial cylindrical ionization chambers listed in IAEA TRS-398. The agreement between the analytical results and the Monte Carlo results was good, whereas the agreement with the recommended value of 0.75 times the cavity radius was good for most ionization chamber but can be different by up to 1.5 mm for some ionization chamber types. In general the recommendation of the IAEA is in good agreement for Farmer type chambers and for other chamber types with a thin wall and a central electrode diameter that is small compared to the cavity diameter. Furthermore, it was found that for ionization chamber with a ratio of central electrode diameter and cavity diameter smaller than 0.25 and a radius smaller than 4 mm and for proton energies above 150 MeV, a simple shift of the entire depth-dose curve is sufficient in order for the difference with the correct curve to be smaller than 3% (the difference being mainly in the Bragg peak).

A modification to the analytical model, dividing the integrations in a number of sub steps, allowed correcting depth-dose curves measured with a cylindrical ionization chamber even in the strongly non-linear gradient region of the Bragg peak. Agreement with the Monte Carlo simulations was again good. This model also enabled to solve the inverse problem of deriving a depth-dose curve in homogeneous water from the depth-dose response obtained with a cylindrical ionization chamber, even from rather noisy data. This inverse optimisation was applied to experimental results from a few papers from the literature and gave satisfactory results although more (and more detailed) data would be welcomed to support this.

G.6. Appendix: Analytical model

The detailed derivations and expressions of the analytical model are described elsewhere (Palmans 2006) but a brief description is given here. The model calculates the water equivalent depth the proton has travelled to when entering the cavity at points Q and T in figure G.6. By integration over the lateral dimensions of the air cavity, a weighted average of the dose to the air cavity at this water equivalent depth is then calculated with the track lengths |QS| and |TU| as weights of the contribution to the total dose. Obviously, if the lateral distance to the z-axis is larger than the central electrode radius, the points S and T

coincide and there is only one contribution with the water equivalent depth of Q and a weight $|QU|$.

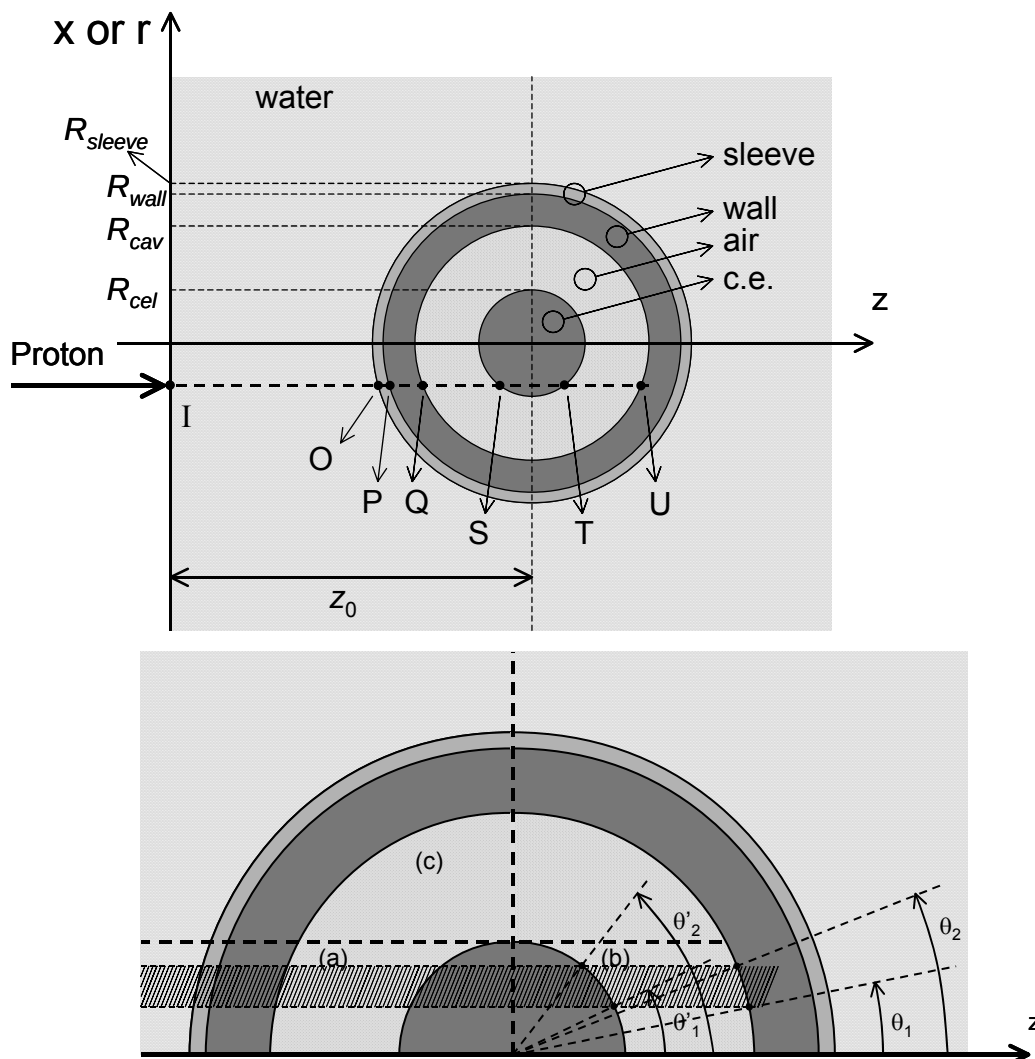


Figure G.6. Upper drawing: transverse cross section through a cylindrical or spherical ionization chamber model for the analytical calculations of the effective depth in water. Lower drawing: detail of the chamber model indicating a subinterval for the integration of $D_{air}(z)$ according to the model of section G.6.2. and definition of the angles θ_1 , θ_2 , θ'_1 and θ'_2 .

The model requires integrals of the following form:

$$\begin{aligned}
 IP^c(R_a, R_b, \theta_1, \theta_2) &= 2 \cdot \int_{x=+R_a \cdot \sin \theta_1}^{x=+R_a \cdot \sin \theta_2} \sqrt{(R_a^2 - x^2) \cdot (R_b^2 - x^2)} \cdot dx \\
 JP^c(R_a, \theta_1, \theta_2) &= 2 \cdot \int_{x=+R_a \cdot \sin \theta_1}^{x=+R_a \cdot \sin \theta_2} \sqrt{(R_a^2 - x^2)} \cdot dx \\
 KP^c(R_a, \theta_1, \theta_2) &= 2 \cdot \int_{x=+R_a \cdot \sin \theta_1}^{x=+R_a \cdot \sin \theta_2} (R_a^2 - x^2) \cdot dx
 \end{aligned} \tag{G.1}$$

for a cylindrical cavity and

$$\begin{aligned}
 IP^s(R_a, R_b, \theta_1, \theta_2) &= \int_{r=R_a \cdot \sin \theta_1}^{r=R_a \cdot \sin \theta_2} \sqrt{(R_a^2 - r^2) \cdot (R_b^2 - r^2)} \cdot r \cdot dr \\
 JP^s(R_a, \theta_1, \theta_2) &= \int_{r=R_a \cdot \sin \theta_1}^{r=R_a \cdot \sin \theta_2} \sqrt{(R_a^2 - r^2)} \cdot r \cdot dr \\
 KP^s(R_a, \theta_1, \theta_2) &= \int_{r=R_a \cdot \sin \theta_1}^{r=R_a \cdot \sin \theta_2} (R_a^2 - r^2) \cdot r \cdot dr
 \end{aligned} \tag{G.2}$$

for a spherical cavity (with $R_a < R_b$) and also the special cases

$$\begin{aligned}
 I^t(R_a, R_b) &= IP^t\left(R_a, R_b, 0, \frac{\pi}{2}\right) \\
 J^t(R_a) &= JP^t\left(R_a, 0, \frac{\pi}{2}\right) \\
 K^t(R_a) &= KP^t\left(R_a, 0, \frac{\pi}{2}\right)
 \end{aligned} \tag{G.3}$$

where the index t refers to the type of cavity (c for cylindrical or s for spherical). All the integrals, except $IP^c(R_a, R_b, \theta_1, \theta_2)$, can be reduced to an analytically closed format.

G.6.1. Analytical model in a linear depth-dose gradient region

In regions where the depth-dose gradient can be considered linear on the local dimensions of the air cavity, a single effective water equivalent depth can be calculated as

$$z_{w,tot}^t = z_0 + \Delta z_{w,tot}^t$$

where the total shift of the water equivalent depth relative to the centre of the cavity, $\Delta z_{w,tot}^t$, is the sum of four contributions:

$$(i) \text{ for the cavity: } \Delta z_{w,cav}^t = -\frac{K^t(R_{cav})}{J^t(R_{cav})} \quad (G.4)$$

$$(ii) \text{ for the wall: } \Delta z_{w,wall}^t = (F_{wall} - 1) \cdot \frac{I^t(R_{cav}, R_{wall}) - K^t(R_{cav})}{J^t(R_{cav})} \quad (G.5)$$

(iii) for the central electrode:

$$\begin{aligned} \Delta z_{w,cel}^t = & \frac{(F_{wall} - 1) \cdot I^t(R_{cav}, R_{wall}) - F_{wall} \cdot K^t(R_{cav})}{J^t(R_{cav}) - J^t(R_{cel})} \cdot \frac{J^t(R_{cel})}{J^t(R_{cav})} \\ & - \frac{(F_{wall} - 1) \cdot I^t(R_{cel}, R_{wall}) - F_{wall} \cdot I^t(R_{cel}, R_{cav})}{J^t(R_{cav}) - J^t(R_{cel})} \\ & + F_{cel} \cdot \frac{I^t(R_{cel}, R_{cav}) - K^t(R_{cel})}{J^t(R_{cav}) - J^t(R_{cel})} \end{aligned} \quad (G.6)$$

(iv) for the sleeve:

$$\begin{aligned} \Delta z_{w,sleeve}^t = & (F_{sleeve} - 1) \cdot \frac{I^t(R_{cav}, R_{sleeve}) - I^t(R_{cav}, R_{wall})}{J^t(R_{cav}) - J^t(R_{cel})} \\ & - (F_{sleeve} - 1) \cdot \frac{I^t(R_{cel}, R_{sleeve}) - I^t(R_{cel}, R_{wall})}{J^t(R_{cav}) - J^t(R_{cel})} \end{aligned} \quad (G.7)$$

It can easily be shown that the equations (G.4) and (G.5) reduce to the equations of Palmans and Verhaegen (1998) for cylindrical and spherical cavities with a non-water-equivalent wall.

It must be remarked that the expressions in equations (G.4-G.7) depend on the order of separating the contributions because the effects are entangled. As a consequence, different results would be found if for example the central electrode and the sleeve effects are calculated in reverse order, or if the central electrode effect would be calculated before the wall effect.

For a realistic thimble-shaped ionization IC ionization chamber the effective water depth is derived as a linear combination of the results for a spherical cavity and for a cylindrical cavity according to the volume fractions of the hemi-spherical and cylindrical parts.

G.6.2. Analytical model in a non-linear depth-dose gradient region

In the presence of a substantial non-linearity of the gradient, such as in the vicinity of the Bragg peak, the model described above is not accurate. The integration should then in

principle be performed with the dose to air as a function of z explicitly in the integrand. But still, a simplified analytical model is possible by dividing the integration interval in a number of subintervals.

Figure G.6 illustrates the integration over a subinterval and defines angles θ_1 , θ_2 , θ'_1 and θ'_2 . With θ_{cel} as the angle defined by the point in the lower graph of figure G.6 where the horizontal dashed line tangent to the central electrode intersects the inner cavity wall, three types of sectors can be indicated: (a) those with $\theta_2 < \theta_{cel}$ and upstream of the central electrode, (b) those with $\theta_2 < \theta_{cel}$ and downstream of the central electrode and (c) those with $\theta_1 > \theta_{cel}$. For simplicity, the case in which $\theta_1 < \theta_{cel} < \theta_2$ is not taken under consideration as it can be reduced to the situations (a), (b) and (c) by introducing an additional subinterval. For every sector the water equivalent depth $z_{w,tot}^t(\theta_1, \theta_2)$ can be calculated by the following expressions:

$$\begin{aligned}
 z_{w,i}^{t,(a)}(\theta_1, \theta_2) = z_0 & - \frac{IP^t(R_{cav}, R_{sleeve}, \theta_1, \theta_2) - IP^t(R_{cel}, R_{sleeve}, \theta'_1, \theta'_2)}{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)} \\
 & + F_{sleeve} \cdot \frac{IP^t(R_{cav}, R_{sleeve}, \theta_1, \theta_2) - IP^t(R_{cel}, R_{sleeve}, \theta'_1, \theta'_2)}{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)} \\
 & - F_{sleeve} \cdot \frac{IP^t(R_{cav}, R_{wall}, \theta_1, \theta_2) - IP^t(R_{cel}, R_{wall}, \theta'_1, \theta'_2)}{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)} \\
 & + F_{wall} \cdot \frac{IP^t(R_{cav}, R_{wall}, \theta_1, \theta_2) - IP^t(R_{cel}, R_{wall}, \theta'_1, \theta'_2)}{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)} \\
 & - F_{wall} \cdot \frac{KP^t(R_{cav}, \theta_1, \theta_2) - IP^t(R_{cel}, R_{cav}, \theta'_1, \theta'_2)}{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)}
 \end{aligned} \tag{G.8}$$

$$z_{w,i}^{t,(b)}(\theta_1, \theta_2) = z_{w,tot}^{t,(a)}(\theta_1, \theta_2) + 2 \cdot F_{cel} \cdot \frac{IP^t(R_{cel}, R_{cav}, \theta'_1, \theta'_2) - KP^t(R_{cel}, \theta'_1, \theta'_2)}{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)} \tag{G.9}$$

$$\begin{aligned}
 z_{w,i}^{t,(c)}(\theta_1, \theta_2) = z_0 & - \frac{IP^t(R_{cav}, R_{sleeve}, \theta_1, \theta_2)}{JP^t(R_{cav}, \theta_1, \theta_2)} \\
 & + F_{sleeve} \cdot \frac{IP^t(R_{cav}, R_{sleeve}, \theta_1, \theta_2) - IP^t(R_{cav}, R_{wall}, \theta_1, \theta_2)}{JP^t(R_{cav}, \theta_1, \theta_2)} \\
 & + F_{wall} \cdot \frac{IP^t(R_{cav}, R_{wall}, \theta_1, \theta_2) - KP^t(R_{cav}, \theta_1, \theta_2)}{JP^t(R_{cav}, \theta_1, \theta_2)}
 \end{aligned} \tag{G.10}$$

The dose measured by the cavity is then again derived by combining the dose for the different sections with the appropriate weight of the sector, which equals the volume fraction of that sector:

$$D_{ch}^t = \sum_i \gamma_i^t \cdot D(z_{w,i}^t) \tag{G.11}$$

where the volume fractions are:

$$\gamma_i^{t,(a)} = \gamma_i^{t,(b)} = \frac{1}{2} \cdot \frac{JP^t(R_{cav}, \theta_1, \theta_2) - JP^t(R_{cel}, \theta'_1, \theta'_2)}{J^t(R_{cav}) - J^t(R_{cel})} \quad (G.12)$$

$$\gamma_i^{t,(c)} = \frac{JP^t(R_1, \theta_1, \theta_2)}{J^t(R_{cav}) - J^t(R_{cel})}$$

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