# Radiation Dosimetry in Medicine: State of the Art in 2007

Proceedings Fifth NCS Lustrum Symposium

**NEDERLANDSE COMMISSIE VOOR STRALINGSDOSIMETRIE**

**Netherlands Commission on Radiation Dosimetry**

**October 2007**

## Preface

The Nederlandse Commissie voor Stralingsdosimetrie (NCS, Netherlands Commission on Radiation Dosimetry, http://ncs-dos.org) was officially established on 3 September 1982 with the aim of promoting the appropriate use of dosimetry of ionizing radiation both for scientific research and practical applications. The NCS is chaired by a board of scientists, installed upon the suggestion of the supporting societies, including the Nederlandse Vereniging voor Radiotherapie en Oncologie (Netherlands Society for Radiotherapy and Oncology), the Nederlandse Vereniging voor Nucleaire Geneeskunde (Dutch Society for Nuclear Medicine), the Nederlandse Vereniging voor Klinische Fysica (Dutch Society for Medical Physics), the Nederlandse Vereniging voor Radiobiologie (Netherlands Radiobiological Society), the Nederlandse Vereniging voor Stralingshygiëne (Netherlands Society for Radiological Protection), the Nederlandse Vereniging voor Medische Beeldvorming en Radiotherapie (Dutch Society for Medical Imaging and Radiotherapy), the Nederlandse Vereniging voor Radiologie (Radiological Society of the Netherlands) and the Belgische Vereniging voor Ziekenhuisfysici/Société Belge des Physiciens des Hôpitaux (Belgian Hospital Physicists Association).

To pursue its aims, the NCS accomplishes the following tasks: participation in dosimetry standardisation and promotion of dosimetry intercomparisons, drafting of dosimetry protocols, collection and evaluation of physical data related to dosimetry. Furthermore the commission shall maintain or establish links with national and international organisations concerned with ionizing radiation and promulgate information on new developments in the field of radiation dosimetry.

# Radiation dosimetry in Medicine: State of the Art in 2007

## Proceedings Fifth NCS Lustrum Symposium

Leiden, The Netherlands, October 19, 2007


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# Foreword - Radiation dosimetry in Medicine: State of the art in 2007.

S. Vynckier
Chairman NCS

Concepts related to dosimetry have a long history; dosimetry plays an important role ever since the discovery of X-rays in 1895 by Wilhelm Rœntgen. On the other hand, one must recognize that developments during the last 10 years have brought the use of ionizing radiation within the hospital to a level of high accuracy and advanced technology. Therefore, at present, dosimetry plays an even more essential role in all clinical processes involving ionizing radiation. The history of the Netherlands Commission on Radiation Dosimetry (NCS: Nederlandse Commissie voor Stralingsdosimetry) covers only two and a half decades, but looking at the various achievements, 25 years of existence are worth to be celebrated. At the occasion of the fifth lustrum, the NCS organizes a symposium in Leiden with the theme “Radiation dosimetry in Medicine: State of art in 2007”.

The NCS lustra were not always celebrated at 5 years intervals. In the spring of 1988 the first lustrum was commemorated with a symposium on thermoluminescent dosimetry at Bilthoven. The lectures were summarized in NCS report 3. The second lustrum was celebrated at Nijmegen in April 1993 on the subject “The role of radiation for detection and treatment of mammary carcinoma”. The third lustrum took place at Leiden in September 1997, on the subject “Dosimetry in and around the hospital”. The proceedings were published in the journal “Klinische Fysica”, issue 1997/2. The fourth lustrum was organized 5 years ago in Delft on the subject “Dosimetry, Perpetual alertness, especially in the digital era”. The activities off all sub-committees were highlighted during this symposium. Again the proceedings were published in the journal “Klinische Fysica”, issue 2002/2+3.

In 1982, when the NCS was established, several objectives were formulated. These goals are still applicable and can be found at the NCS website: http://www.ncs-dos.org:

- Participation in dosimetry standardization and promotion of dosimetry intercomparisons;
- Drafting of dosimetry protocols;
- Collection and evaluation of physical data related to radiation dosimetry;
- Maintain or establish links with national and international organizations concerned with ionizing radiation;
- Promulgate information on new developments in the field of radiation dosimetry.

Since the installation of the NCS in 1982, different sub-committees were formed to fulfil these objectives. Their achievements were published in a total of 17 NCS reports:

- Radiation dosimetry activities in the Netherlands. NCS Report 1, July 1986
- Code of practice for the dosimetry of high-energy photon beams. NCS Report 2, December 1986.
- Proceedings of the symposium on thermoluminescent dosimetry. NCS Report 3, October 1988.
- Recommendations for dosimetry and quality control of radioactive sources used in brachytherapy. NCS Report 4, February 1989.
- Code of practice for the dosimetry of high-energy electron beams. NCS Report 5, December 1989.
- Dosimetry aspects of mammography. NCS Report 6, March 1993.
- Recommendations for the calibration of iridium-192 high dose rate sources. NCS report 7, December 1994.
- Kwaliteitscontrole van medisch lineaire versnellers: methoden voor kwaliteitscontrole, wenselijke toleranties en frequenties. NCS Report 8, December 1995.
- Quality control of medical linear accelerators: current practice and minimum requirements. NCS Report 9 August 1996.
- Dosimetry of low and medium energy X-rays, a code of practice for use in radiotherapy and radiobiology. NCS Report 10, July 1997.
- Quality control (QC) of simulators and CT-scanners and some basic QC methods for treatment planning systems: current practice and minimum requirements. NCS Report 11, September 1997.
- Determination and use of scatter correction factors of megavoltage photon beams. NCS Report 12, March 1998.
- Quality Control in brachytherapy: current practice and minimum requirements. NCS Report 13, November 2000.
- Quality control of sealed beta sources in brachytherapy: recommendations on detectors, measurement procedures and quality control of beta sources. NCS Report 14, 2005.
- Quality assurance of 3-D treatment planning systems for external photon and electron beams: practical guidelines for acceptance testing, commissioning and periodic quality control of radiation therapy treatment planning systems. NCS Report 15, 2006.
- Monte Carlo treatment planning, an introduction. NCS Report 16, 2006.

- Dosimetrie in de radiologie: stralingsbelasting van de patiënt en werknemers. NCS Report 17, 2007.

It is gratifying to see that most of these reports have led to a better understanding of several dosimetry issues and moreover have promoted the uniformity in dosimetry procedures applied in hospitals. Furthermore, these reports are regularly quoted in the international literature.

The subject of the fourth lustrum symposium of the NCS was: “Dosimetry, a perpetual alertness, especially in the digital era”. The goal of this symposium was to keep the awareness of medical physicists, performing dosimetry in a hospital environment, at a high level. I have started this text with a sentence on history, but we could also ask: where are we now in 2007? The NCS board has taken the initiative to organize the fifth lustrum in Leiden with the title: “Radiation dosimetry in Medicine: State of art in 2007.” This fifth lustrum will be an opportunity for the different sub-committees of the NCS to present the progress made in their research areas. Presentations will be given by the representatives of the following subcommittees:

- Uniformity of dosimetry protocols;
- Dosimetry in radiology;
- Quality assurance of brachytherapy systems;
- Reference values in radiology;
- Monte Carlo treatment planning;
- Film dosimetry and other issues.

As chairman of the NCS, I must gratefully acknowledge the scientists in the Netherlands and Belgium, involved in the activities of the different NCS subcommittees. Most of their work is done on a voluntary basis, but results in high quality reports that are consulted on a national and international level. Together with the board, I shall make the effort to maintain this quality standard.

# Clinical reference dosimetry based on absorbed dose to water standards: a new NCS code of practice for external beam therapy

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## Introduction

Accurate determination of the dose delivered to the tumour in external beam radiotherapy is of primary importance in clinical dosimetry. Until recently, the determination of absorbed dose under reference conditions has been based on the use of ionisation chambers calibrated in terms of air kerma as recommended in codes of practice, such as published in NCS Reports 2 and 5 [1,2]. An important reason to issue these NCS dosimetry protocols was the revision of reference data and physical parameters as recommended by international organisations CCEMRI (1985) and ICRU (1984) [3,4]. The adoption of these revised data not only influenced the absorbed dose determination in clinical practice but also required corresponding changes in the exposure and air kerma standards in national metrology institutes worldwide. As a result of these developments the primary dosimetry standards for photon radiation in the Netherlands were revised in 1987.

The NCS codes of practice for the dosimetry in high energy photon and electron beams were based on the concept of using single conversion factors to convert the reading of an ionisation chamber to absorbed dose to water as a function of radiation quality. These codes of practice were kept brief and relatively simple and correspond to the current clinical practice in Belgium and The Netherlands, but the underlying physics: equations and numerical data for physical parameters, correction and conversion factors were provided in appendices. The recommendations given in the NCS codes of practice are applied in every radiotherapy institute in Belgium and the Netherlands since 1986.

Although these protocols based on air kerma in $^{60}$Co gamma radiation meant a significant step forward several problems remained. These protocols were inherently complex reducing the overall accuracy of dosimetry. The complexity arose mainly from the fact that the ionisation chamber had to be calibrated free-in-air for the quantity air kerma which had to be converted to obtain the quantity absorbed dose to water involving a measurement in a phantom. Furthermore the air kerma standards operated by the national standard laboratories were based on the same measurement technique and therefore subject to common errors. To overcome these

disadvantages many standard laboratories developed measurement standards of absorbed dose to water.

**Absorbed dose standards**

A significant advantage of these absorbed dose standards is that they are based on different techniques such as graphite and water calorimeters, energy absorption in a Fricke solution and ionometric methods. The most direct technique is based on water calorimetry, either by using a small volume of sealed, high-purity water in a water phantom to perform a direct measurement of absorbed dose to water at a point or by calibrating a Fricke solution in the beam quality of interest. A second technique based on a graphite calorimeter is a two-step approach. In the first step the absorbed dose to graphite is established and in the second step various procedures are used to determine the absorbed dose to water in the same beam. A third technique is based on the total absorption in a Fricke solution of an electron beam of known energy. The total absorption of the electrons is used to calibrate the solution and then the Fricke solution is placed in a small glass ampoule to establish the dose at a point in a water phantom. It has been assumed in the past that the calibration of the Fricke solution does not change with beam quality. More recently this assumption came under scrutiny when it was demonstrated that the sensitivity of a Fricke solution varies by about 0,7% over clinical photon beam range. The ionometric method is used by the BIPM by directly measuring the absorbed dose to water in a water phantom employing a water-tight graphite-cavity ionisation chamber with an accurately known volume.

At present the majority of primary standards laboratories have based their absorbed dose standards either on a graphite calorimeter or on a water calorimeter. Primary standards laboratories have set up calibration services for measuring absorbed dose to water in $^{60}$Co beams or as a function of different beam qualities in accelerator beams. Comparisons of primary absorbed dose standards have been conducted over the years based on the various methods described above. The results of these comparisons show a satisfactory agreement at the 1% level or better. It is important to note that each of the methods involved has different types of uncertainties (Type B) and thus the international system of absorbed dose standards is considered a robust system with respect to the Type B uncertainties affecting all standards.

Advances made in radiation dosimetry concepts and the availability of absorbed dose standards prompted the development of new codes of practice for clinical reference dosimetry in high energy photon and electron beams based on absorbed dose to water standards. These new protocols are replacing the air kerma based protocols applied for the last twenty years.

**The new NCS code of practice for external beam therapy**

The international developments made in radiation dosimetry during the last decades and the publication of new dosimetry protocols based on absorbed dose to water standards by task group 51 of the AAPM [5] and by the IAEA (TRS-398) [6] were strong incentives to revise the NCS codes of practice based on air kerma. The NCS subcommittee on “Uniformity Dosimetry Protocols” was revived and has drafted a code of practice based on the absorbed dose to water concept for the reference dosimetry in clinical high energy photon beams with nominal acceleration potentials between 1 and 25 MV and high energy electron beams with nominal energies between 4 and 25 MeV. The NCS code is based on absorbed dose to water standards for $^{60}$Co reference beams. The national standards laboratories in Belgium and The Netherlands, the Laboratory for Standard Dosimetry Ghent (LSDG) and the Netherlands Measurements Institute (NMi) respectively, operate calibration services for absorbed dose to water in $^{60}$Co gamma radiation. The primary standards involved are based on sealed water calorimeters according to a design of Domen [7]. A more detailed description of these standards can be found elsewhere [8,9].

In this new code of practice a limited number of ionisation chambers for reference dosimetry is recommended, but the physical concepts outlined in the code represent a major simplification compared to the previous NCS codes based on the concepts of air kerma employing Bragg-Gray or Spencer-Attix theory. Furthermore this code of practice introduces a single beam quality correction factor taking into account all effects dependent of the radiation beam quality. This beam quality correction factor is defined as the ratio of absorbed dose to water calibration coefficients in the clinical beam quality $Q$ and the reference beam quality $Q_0$:

$$k_{Q,Q_0} = \frac{N_{D,w,Q}}{N_{D,w,Q_0}}$$

The reference beam quality $Q_0$ used for the calibration of ionisation chambers is $^{60}$Co gamma radiation. This approach leads to a basic formalism for the absorbed dose determination, which is very similar to the one used in the previous NCS codes of practice based on air kerma. The absorbed dose to water at the reference depth in water for a beam quality $Q$ in absence of the ionisation chamber is given by:

$$D_{w,Q} = M_{\mathrm{corr},Q}\, N_{D,w}\, k_{Q,Qo}$$

where :

$M_{corr,Q}$ reading of the electrometer corrected to ambient reference conditions and for the effects of recombination, polarity and the influence of the electrometer,

$N_{D,w}$ the calibration coefficient for absorbed dose to water in $^{60}$Co gamma radiation,

$k_{Q,Qo}$ beam quality correction factor.

For photon beams the beam quality correction factors are based on experimental data. $k_{Q,Qo}$ values were measured in selected clinical high-energy photon beams in Belgium and The Netherlands together with extensive measurements of the beam quality specifiers $TPR_{20,10}$ and $\%dd(10)_x$. The measurements involved the use of a portable water calorimeter and were performed for four types of graphite walled cylindrical ionisation chambers (NE2611A, NE2571, PTW 30012 and Wellhöfer FC65G). In Figure 1 a typical setup of the portable water calorimeter in a clinical accelerator beam is shown. The portable water calorimeter was constructed at NMi and extensively tested in $^{60}$Co gamma radiation and in various photon beams produced by medical accelerators.

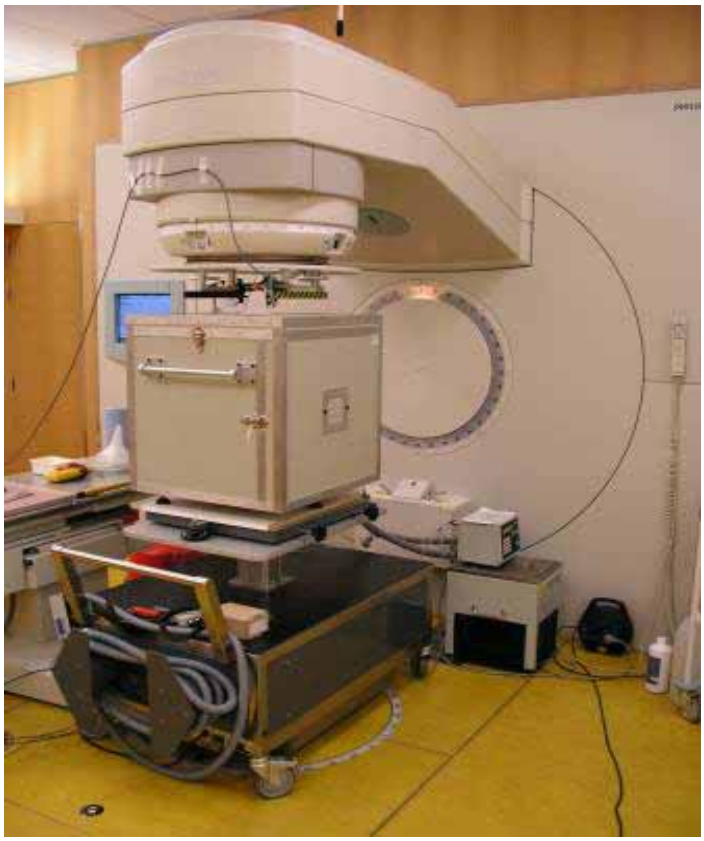

*Figure 1: Setup of the portable NMi water calorimeter in a clinical photon beam*

These measured $k_{Q,Qo}$ values are combined with existing and new experimental data in a model analysis with a thorough uncertainty evaluation proving that there is no obvious advantage in using either of the beam quality specifiers and providing an easy parametric representation of the data which is convenient to use in clinical practice.
In the code of practice the $k_{Q,Qo}$ data for photon beams are given as a function of beam quality specifier $TPR_{20,10}$ and are represented in the form of sigmoid fits. $TPR_{20,10}$ is preferred as beam quality specifier for high energy photon beams, due to the fact that it is independent of the electron contamination in the incident beam. It is also a measure of the effective attenuation coefficient describing the approximately exponential decrease of a photon depth-dose curve beyond the depth of maximum dose. Furthermore it does not require the use of displacement correction factors at two depths when cylindrical chambers are used and is therefore practically not affected by systematic errors in positioning the chamber at each depth, as the settings in the two positions will be affected in a similar manner. Finally, it is a parameter that is normally available through commissioning measurements and does not require special measurements using lead filters as it is the case for the $\%dd(10)_x$ parameter.

In Figures 2A and 2B examples of the sigmoid model for the $k_{Q,Qo}$ data for photon beams are given. All data points for respectively the NE2571 and NE2611A ionisation chamber are shown as a function of $TPR_{20,10}$ together with the sigmoid fit. Theoretical data from TRS-398 are shown as well.

For electron beams theoretically calculated beam quality correction factors, based on recent literature data, are given as function of the half-value depth in water $R_{50,dos}$. These factors are given for the Farmer-type graphite walled cylindrical ionisation chambers recommended for reference dosimetry in photon beams and for the designated NACP02 and Roos type plane-parallel ionisation chambers. The choice of $R_{50,dos}$ results in a simplification of the procedures in NCS report 5 involving the derivation of the mean energy at the phantom surface based on $R_{50,dos}$ as an intermediate step. It also directly relates the beam quality specifier to the penetration characteristics of the electron beam. The definition of $R_{50,dos}$ is made in terms of absorbed dose levels, whereas usually ionisation curves are measured. Apart from the introduction of $R_{50,dos}$ as beam quality specifier, the reference depth for electron dosimetry has changed compared to NCS-report 5, based on new stopping power calculations for clinical electron beams.

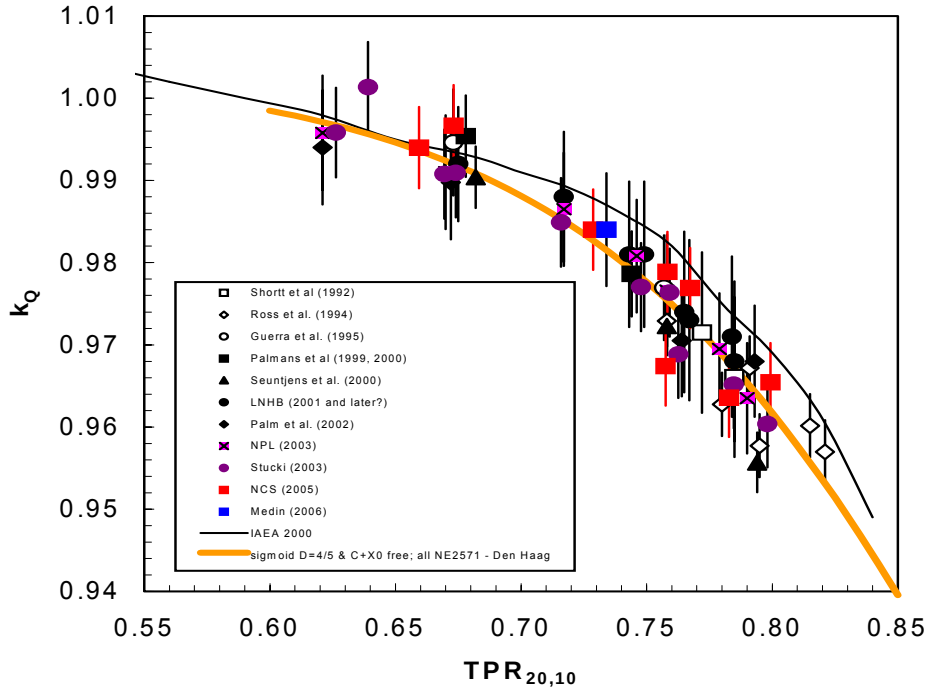


*Figure 2A: $k_{Q,Q_o}$ data from the literature together with the measured NCS data for the NE 2571 type ionisation chamber as a function of $TPR_{20,10}$ with a sigmoid fit.*

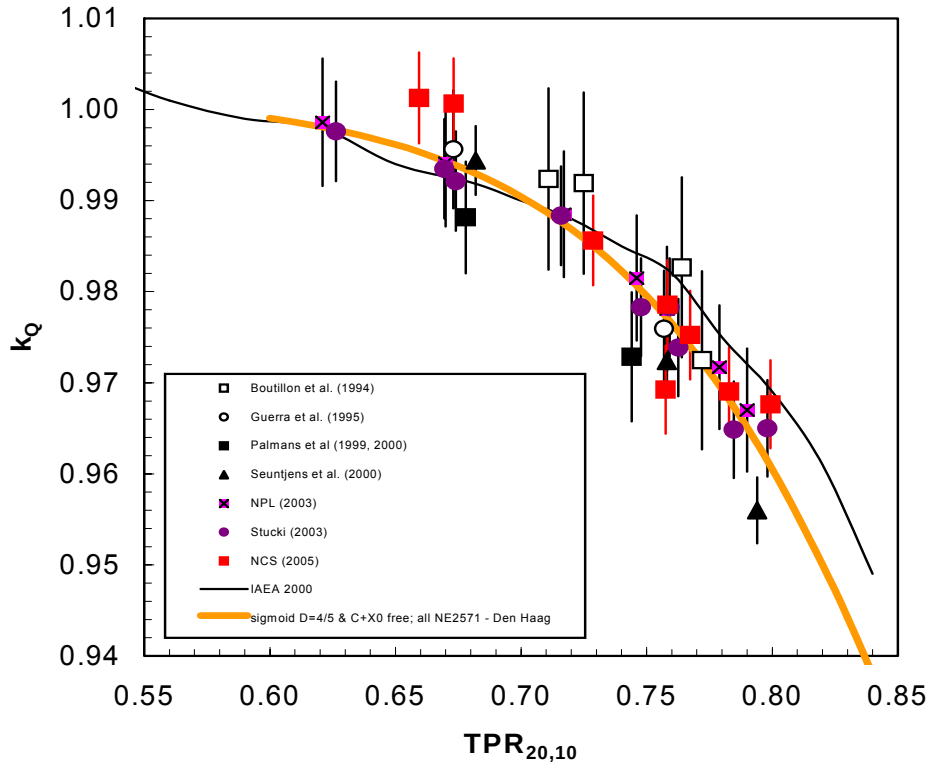


*Figure 2B: $k_{Q,Q_o}$ data from the literature together with the measured NCS data for the NE 2611A type ionisation chamber as a function of $TPR_{20,10}$ with a sigmoid fit.*

The new reference depth $z_{ref}$ is defined by:

$$z_{ref} = 0.6\, R_{50,dos} - 0.1 \text{ cm}$$

For high-energy electron beams, new experimental data on $p_{wall}$ for plane-parallel ionisation chambers in the $^{60}$Co calibration beam became available as well as new Monte Carlo calculated values of the overall perturbation factors $p_Q$ for plane-parallel ionisation chambers. The new data for the perturbation factors has been used in the determination of values for the beam quality correction factors for plane-parallel chambers as a function of quality index $R_{50,dos}$.

The new code of practice contains a number of appendices giving information on recommended ionisation chambers, on methods, physical and numerical data concerning influence quantities, on absorbed dose standards and beam quality correction factors and on the differences between the new code of practice and the previous NCS codes. A special appendix is devoted to the estimation of uncertainty according to international guidelines given by ISO and EA [10,11].

**Future developments**

The code of practice covers the reference dosimetry in photon and electron beams produced by conventional linear accelerators. However, in external radiotherapy a strong growth is observed in various treatment modalities, which depends strongly on small treatment fields such as intensity modulated radiotherapy (IMRT), image guided radiotherapy (IGRT), stereotactic radiotherapy, robotic radiotherapy and tomotherapy. Also an growing interest is expected in charged particle (hadron) therapy, in particular the use of proton beams.

Furthermore the increased use of imaging techniques in the verification and adjustment of the dose delivered to patients in radiotherapy, the application of time dependent data of patient movement (e.g. in IGRT and 4D therapy) will lead to the development of computer controlled dynamic irradiation facilities governed by accurate treatment planning systems based on Monte Carlo driven “dose-engines”, which dynamically plan, treat, adjust and verify the dose to the patient.

These rapid evolving radiotherapy techniques require the development of novel (e.g. 3D) dosimetry techniques, dose mapping methods, new dosimetry standards, improved evaluation of interaction data, codes of practice and the definition of new reference conditions to ensure the consistency in radiation dosimetry between conventional radiotherapy and these new treatment techniques and to allow for traceability to primary measurement standards.

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# Dosimetry in Radiology: radiation burden to patient and workers

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**Abstract**

The NCS has recently issued a report on dosimetry in radiology. This publication covers radiation quantities and units for dosimetry for patients as well as staff involved in diagnostic and interventional radiology. Risks of exposure to ionising radiation in diagnostic and interventional radiology are reviewed. Dosimetry methods using active or passive devices are summarised including, calibration of dosimeters. Patient and staff dosimetry are given in the report, as well as methods for assessment of image quality and dose reduction. Backscatter factors are presented separately.

## Introduction

Dosimetry in radiology was stimulated by the European Medical Exposure Directive [1]. Until 1996, dosimetry was restricted to a few university hospitals, but since then more radiology departments perform dosimetry in the framework of risk assessment and quality control. The NCS installed a subcommittee with the aim of harmonising dosimetry in radiology since at the time the quantities used were confusing. The contribution of backscattered radiation was in the dose assessment was not always considered adequately. In 1995 the International Commission on Radiation Units and Measurements established a Report Committee with the aim of harmonising patient dosimetry in radiology. The NCS and ICRU agreed to recommend new quantities for patient dosimetry at approximately the same time. This decision has caused a considerable delay in the publication of the NCS report as the ICRU Report was only produced in 2005 [2]. Since it was anticipated that radiographers and radiological technologists will play a major role in performing dosimetry in radiology the NCS report was written in Dutch.

## From source to recipient

The history of a photon in imaging using x rays can be characterised by a sequence of events from source to image receptor (see Figure 1). The ICRU [3] has defined fundamental quantities and units to describe the processes involved. Important parameters of the x-ray source are the source strength and the radiation quality, related to the tube voltage, the total filtration and the half-value layer. The radiation field can be quantified by the fluence or kerma values. In the NCS report incident air kerma (without backscatter), entrance air kerma (with backscatter) and

air kerma-area product are defined in accordance with [2]. Specific recommendations are provided for CT examinations.

Inside the human body the important quantities are organ dose and equivalent organ dose. In interventional radiology, special attention has to be given to probable high doses in localised regions of the body, usually the skin. For purposes of radiological protection, effective dose is the most suitable quantity for risk assessment due to exposure to ionising radiation [4]. On the basis of input parameters such as incident air kerma, organ doses can be calculated with the aid of radiation transport simulations using Monte Carlo methods. Subsequently, the effective dose can be derived as the summation of the products of organ dose and tissue weighting factors. Recently, more detailed information became available on tissue weighting factors for relevant organs and tissues [5,6].

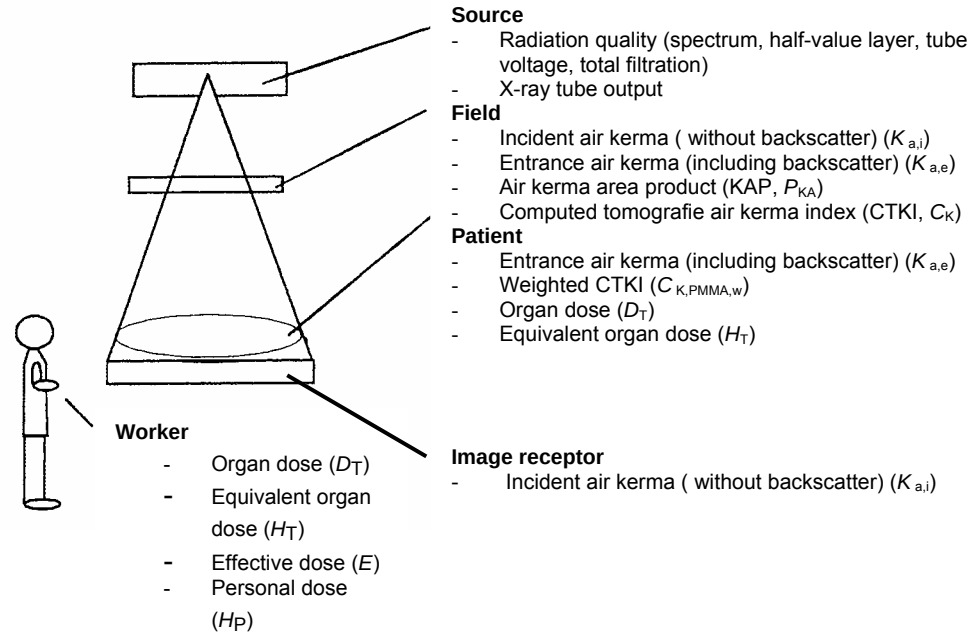


*Figure 1: Radiation quantities in radiology, related to the pathway of the photons from source to image receptor, including the patient and the radiological worker*

In practice the effective dose cannot be easily determined for radiological workers. The quantity personal dose equivalent can be measured with personal dosimeters and is expressed in soft tissue at a depth under the positions of the dosimeter. For photons of sufficiently high energy a depth of 10 mm is used. The application of the personal dose is based upon the assumption that this gives an overestimation of the effective dose.

**Risks of exposure to ionising radiation**

The detrimental consequences of radiation exposure can be distinguished in deterministic effects and stochastic effects. For deterministic effects, the probability and the severity of the effect are dependent on the dose. The deterministic effects are called tissue reactions in the new ICRP recommendations [5] and will only become clinically manifest when a threshold dose is exceeded. The tissue reactions and threshold doses are summarised for a number of tissues including the skin.

For stochastic effects only the probability of occurrence is dependent on the dose and its administration in the course of time. Tumour induction is the most important stochastic effect. For this process there is no evidence of a threshold dose. The risk for induction of fatal malignancies is strongly dependent on the age of the individual at the time of irradiation. For exposure to high dose and high dose rates the life time risk for the total population is approximately 10 percent per Sievert (Sv). In diagnostic radiology and radiological protection the doses and dose rates are usually limited. Applying a dose and dose rate effectiveness factor (DDREF) of 2 the risk of induction of fatal cancers is 4.6 percent per Sv for adults and 5.9 percent for the total population.

**Dosimetry methods**

Dosimeters used in radiology and some of their characteristics are:

- For measurement of air kerma for the relatively low energy x-rays [1] used in radiology various types and shapes of ionisation chambers may be used. They are presented in the report as well as corrections to be made to the readings, including environmental pressure and temperature, ion recombination, polarity effect, angular dependence and energy dependence.
- Semi-conductor devices are also active dosimeters. They are favourable in view of their increased sensitivity resulting in smaller sensitive volumes compared to ionisation chambers. The main disadvantage is their relatively large energy dependence.
- Advantages of thermoluminescent detectors (TLD) are their small size and ,e.g., the lack of cables when exposed on the skin of patients. Drawbacks are the energy dependence and that it is a passive method.
- Transmission ionisation chambers can be used to determine air kerma-area product. Advantages are that they can be used at any distance from the focus of the x-ray tube

and that the reading is a measure for air kerma as well as the beam area. A disadvantage may be that *in situ* calibration is necessary.

**Determination of patient dose in radiology**

Organ or local doses can be derived from measurements using anthropomorphic phantoms and Monte Carlo simulations of radiation transport in mathematical or voxel phantoms. During the past decade Dutch scientists have investigated effective doses for relatively complex procedures in radiology. The resulting effective doses (average values and range) are given in Table 1.

*Table 1: Average effective dose and range for relatively complex diagnostic en interventional radiology procedures in the Netherlands.*

| Procedure | Effective dose (mSv) | |
|---|---|---|
| | average | range |
| Neural interventions [7] | 14 | 6 – 22 |
| Vascular interventions [8,9] | 12 | 12 – 13 |
| Digital Subtraction Angiography (DSA) of renal arteries [10] | 9.1 | – |
| DSA of lung vessels [11] | 7.1 | 3 – 17 |
| Oesophagus-stomach [12] | 7 | 3 – 19 |
| Double contrast barium enema [13] | 6.4 | 2 – 10 |
| Intraveneous DSA [8] | 6 | 2 – 10 |
| Upper colon [14] | 5 | 3 – 8 |
| Intra-arterial arteriography [8,9] | 4 | 3 – 8 |
| Intra-oral dental examination [15] | < 0/01 | – |

Diagnostic reference levels (DRL) have been proposed by the International Commission on Radiological Protection (ICRP) [16]. European reference levels have been formulated for various types of diagnostic and interventional procedures, but not up to the present in the Netherlands.

**Individual monitoring, staff dosimetry**

The aims of individual monitoring, include monitoring policy, type and application of personal dosimeters, wearing position(s) in presence or absence of protective clothing and the consequences of the individual monitoring results. The application of protective clothing will have serious consequences concerning the validity of the results of individual monitoring. These are dealt with in the report for different conditions. Various examples of the consequences of individual monitoring for the optimisation of radiological protection are provided.

**Assessment of image quality**

Various methods for assessment of image quality in radiology are available; ranging from fundamental methods, in terms of large transfer function, spatial resolution and noise; statistical decision theory, in terms of various types of observer; psycho-physical approaches, e.g. receiver-operating characteristic (ROC) analysis and contrast-detail analysis; and judgment of the quality of images employing image quality criteria. Various studies on image quality in radiology revealed relatively small variations in image quality whereas large dose variations were observed.

**Conclusion**

Doses to patients as well as workers may be considerable, particularly in interventional radiology**.** For patients, doses and resulting risks have to be weighted against the benefits of the use of ionising radiation. Possibilities are given for dose reduction for patients as well as for staff. Consequences of measures applied for dose reduction on image quality are also provided. It is anticipated that NCS report 17 can be used in practice by radiological technicians as well as clinical physicists for patient and staff dosimetry in radiology.

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# Towards recommendations for quality control of low photon energy emitting seeds in brachytherapy

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NCS Subcommittee: Quality control of low photon energy emitting seeds in brachytherapy

**Background and Aim**

The incidence of prostate cancer has shown a large increase in the last decades. This can be linked to the introduction of specific blood tests (PSA-test), which enabled easy screening and detection of even early stage prostate cancer. As relatively more patients with early stage disease presented themselves for treatment, also a shift in treatment modalities could be observed, and the number of patients treated for prostate cancer with brachytherapy increased considerably. Indeed permanent implant prostate brachytherapy (PPBT) has proven to be a valid treatment option for early stage disease, offering these patients comparable treatment outcome as external beam radiotherapy or surgery, with a favourable site effect profile [1].

This has lead to a vast increase in the use of low photon energy sources in brachytherapy, where actually the use of I-125 sources for PPBT is by far the major application of low-photon-energy emitting sources in Belgium and The Netherlands.
This increased use stimulated the Netherlands Commission on Radiation Dosimetry (NCS) to establish in February 2004 a sub-commission in order to study the clinical practice of quality assurance aspects related to the use of these low photon energy sources. The aim was to make an overview of the currently applied QA procedures in PPBT, and to publish a report with guidelines and recommendations on QC regarding dose calculation and seed calibrations.

The work of this group should also stimulate the development of a standard for such sources in Belgium and The Netherlands, and promote efforts to make calibration methods at each center traceable to (inter)national measurement standards.

**Materials and Methods**

To gain insight in the current practice of QA of seed implantations, a questionnaire was distributed to all radiotherapy institutions in Belgium and The Netherlands. The questions were related to prostate implantation procedures and techniques, treatment planning and source

calibration. Based on these results, it was decided to develop a user test procedure for treatment planning systems (TPS), and to validate the calibration procedure in all institutions.

The TPS test procedure focussed on 5 aspects: absolute dose calculation in a number of points on the central axis of the source (1D anisotropy approximation), dose summation, isodose line representation, dose-volume histogram (DVH) calculation, and absolute dose calculation using the 2D anisotropy correction. As all systems used in our countries are applying the AAPM TG-43 [2],[3] calculation method we expected possible deviations to be linked most probably to differences in TG-43 source reference data, used as input for the TPS. Therefore we also recorded the reference data used in each institution. Whenever possible we used the 2004 AAPM TG-43U1 [3] and 2007 AAPM TG-43U1S [4] consensus source data as reference.

In-air kerma measurements were performed in all participating institutions by a visiting team, using two commercially available measurement systems, both calibrated at the Dutch Standards Laboratory (NMi). These two instruments, the PTW SourceCheck and the Standard Imaging IVB1000 well type chamber, were chosen as they can be equipped with dedicated adaptors allowing measurements performed in a precise and reproducible way for single seeds as well as for seeds in specific clinical packaging (strands-cartridges). The results were compared to local measurements obtained with the centre measuring equipment (if available), and to the source strength specified on the manufacturer's certificate. A NIST traceable calibration coefficient for each seed model/brand used in our countries was available for our equipment.

In order to better understand source strength values and uncertainties as specified by the manufacturers, the sub-commission also studied the manufacturing process, and especially the calibration and QC procedures that are applied by the different manufacturers. In contrast to other brachytherapy applications where often a single source or limited a number of (wire) sources is used, in PPBT a large number of seeds (30-100) are implanted during the procedure. In general manufacturers measure each individual seed, but rather than specifying the strength for each individual seed, they classify the seeds in different groups each containing seeds with air kerma strength within a certain range (binning). Usually these bins are characterized by a nominal source strength, the nominal mid value of the bin. Steps of about 10% (= 1 week decay) between two consecutive bins are often used. In this case the user receives seeds with a value for the strength equal to the nominal value as specified on the certificate ±5% due to the binning process, and ±7% or more when taking into account measurement uncertainties. Some

manufacturers might adapt a more narrow binning range, leading to a somewhat smaller uncertainty.

## Results

### 1 Survey

The questionnaire was first send around in 2004 and updated in January 2006. All centres performing PPBT responded: 22 RT departments in Belgium and 12 RT departments in The Netherlands. Only the use of I-125 sources was reported, most often for PPBT, but also for eye and mamma treatments (each in 1 institute). The different seed models/brands used are presented in table 1.

In prostate BT, the I-125 sources are used as single seeds, strands or in Mick cartridges.

*Table 1: Number of users of different source models in Belgium (Be) and The Netherlands (Nl) at the time of the survey (January 2006).*

| | Oncura | IBt | | Bebig | Bard | Isotron |
|---|---|---|---|---|---|---|
| | Rapidstrand | Intersource | Interstrand | Isocord I25.S171 | STM1251 | Interseed |
| Be | 10 | 7 | 4 | 1 | 1 | - |
| Nl | 5 | 1 | 6 | - | - | 1 |

Four different TPS systems were encountered, of which Variseed (Varian) was most frequently used (table 2). In all TPS systems, AAPM TG-43 based dose formalism and algorithms are implemented. A large variability was observed in the calculation model (point source or line source approximation), and anisotropy correction (constant, 1D factor or 2D function). Nine centres still applied the anisotropy constant as correction factor, while this was no longer recommended by AAPM TG-43U1 since 2004.

*Table 2: Treatment planning systems used for dose calculation in PPBT.*

| VariSeed | Prowess | PSID | Spotpro |
|---|---|---|---|
| 30 | 2 | 2 | 1 |

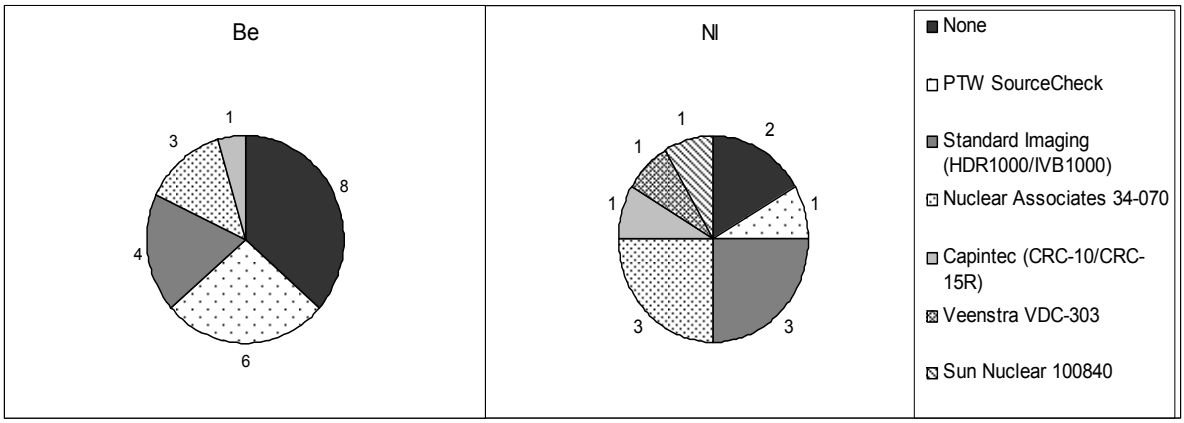


*Figure 1: QC instruments used in the different institutions in Belgium and The Netherlands. Number of instruments of each brand is indicated on the graph.*

There is a large variety of QC instruments and methods for verifying the air kerma strength of the seeds. In total 10 out of the 34 institutions reported not having a specific instrument to verify the source strength of the seeds used for PPBT, and 5 others stated not performing any measurement (yet) in clinical routine.

Of the remaining centres some measure all seeds/strands before use, some take a sample of a few seeds, while others measure some of the remaining seeds after the implantation. Major reasons mentioned for not measuring are the fact that seeds/strands should be handled in sterile conditions, and inappropriate measurement equipment to handle stranded seeds, or seeds in cartridges.

Of the 24 instruments available in the hospitals only 3 were calibrated at SSDL-level (all at an Accredited Dosimetry Calibration Laboratory), 7 were calibrated by the manufacturer (the PTW instruments), while the others had been checked by using “calibrated” seeds obtained from the source manufacturer or were used only as a tool to check the internal consistency.

**2 TPS tests**

Important deviations of up to 11% were observed in the dose calculations at fixed points, even in the simple case of 1D dose calculation (figure 2). These large deviations could not be contributed to an incorrect calculation or wrong implementation of the algorithm, but the use of different TG-43 source datasets showed to have a major impact, some centres using outdated data, others not using a correct implementation of the data and using line source datasets in a point source calculation model or vice versa. Also some deviations were caused because the test guidelines were not exactly followed. It should be noted that we report some systematic deviation for the IBt seed in figure 2, which is due to the fact that we decided to use the AAPM

TG-43U1S consensus dataset as reference for this source which was only published in June 2007 and which could thus not have been implemented by the users at the time of the test.
Similar observations were found for the 2D-dose calculation verification.
No problems were observed with respect to the isodose representation, while the results of the DVH test need to be further analyzed.

### 3 On-site visits

19 centres in Belgium and 12 centres in The Netherlands participated to this study. 3 centres decided not to participate, as they did not routinely perform any verification measurement of the seed strengths. The visiting team started in February 2006 and finished its work by the end of September 2006. About 100 individual seeds were measured, together with some stranded seeds and cartridges. All measurements were within the tolerances as specified by the source manufacturer on the source certificate, also taking into account the measurement uncertainty. The mean ratio of specified strength over measured strength was close to unity (1.004 for Oncura 6711, .999 for IBt Interseed/Interstrand). In the few centres where a calibrated instrument was available our results showed good agreement with the local measurements.

### Conclusion

The number of institutions performing PPBT, and the number of patients treated with this technique have increased dramatically in the last four years. It was anticipated that this increase might have gone along with the implementation of different levels of QA procedures in the different institutions, and our survey, and on-site visits have confirmed this.

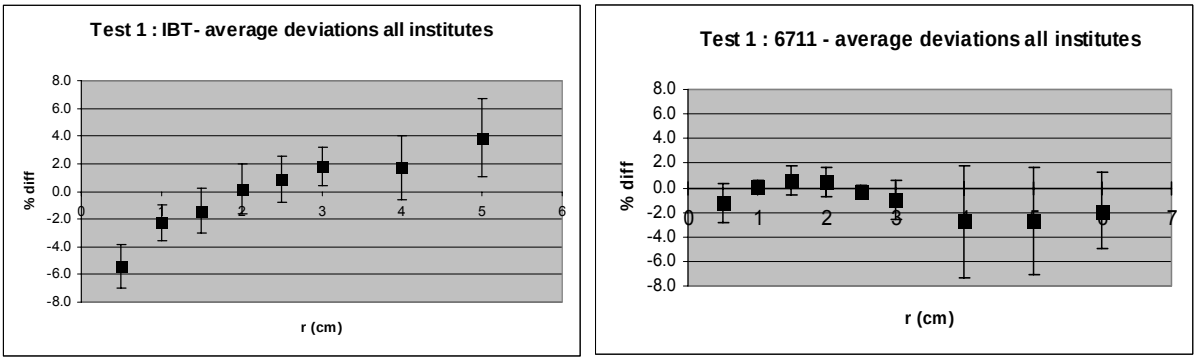


*Figure 2: Results of TPS test 1: absolute dose calculation in points on the source central axis for 2 different sources: IBt Intersource/Interstrand and Oncura 6711/Rapidstrand. The average deviation (± 1 standard deviation) over all institutions using these seeds is indicated at different distances r from the source.*

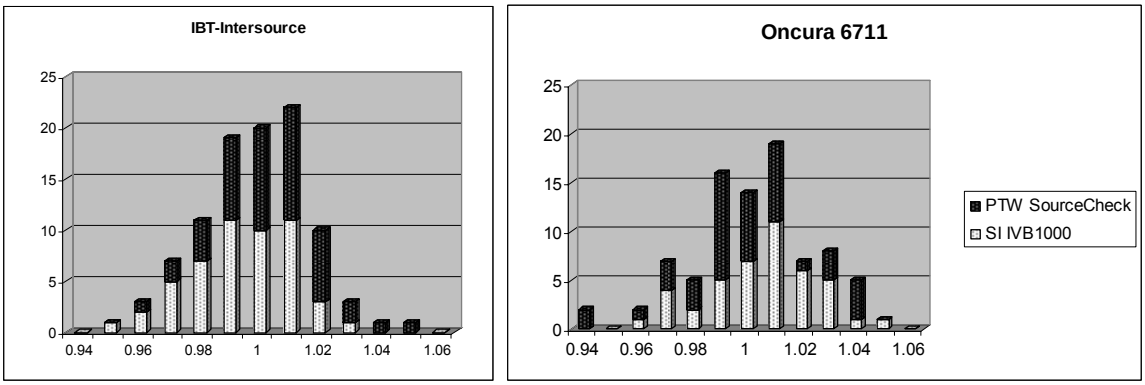


*Figure 3: Results of the on-site visits: histogram of the ratio of the (manufacturer) stated source strength over the (NCS) measured source strength for 2 different sources: IBt Intersource/Interstrand and Oncura 6711/Rapidstrand. Measurement results for the 2 instruments used are shown.*

With respect to source strength verification the lack of (international) recommendations, unawareness on how to maintain sterility during measurement, and difficulty to obtain a traceable calibration factor seem to be the major drawbacks for the physicists, and a large variability was observed in procedures, timing and measurement equipment used.

Most users were found to apply the TG-43-data supplied by the TPS manufacturer, but these data are not always kept up-to-date. Clinical importance of the deviations that have been observed should however be considered as limited, as due to the nature of the procedure and the large number of implanted seeds dose rates at very close distance (.5-2 cm) are predominant for the final dose distribution.

The NCS sub-commission aims at publishing later this year a report with recommendations for QC of low-energy-photon sources, thus making a contribution to improve and harmonize the QC efforts in all hospitals in Belgium and The Netherlands.

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# Monte Carlo treatment planning: theory and practice - status of the activities of the NCS sub-commission on Monte Carlo treatment planning

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NCS Subcommittee: Monte Carlo Treatment Planning

During the last decades, technical possibilities in external beam radiotherapy have developed substantially (IMRT, tomotherapy, robotic therapy, hadron therapy, image guided therapy, gating, …). In addition, important improvements have been made in treatment planning software, including options for inverse planning, and advanced dose calculation algorithms. Originally, dose calculations relied heavily on analytic, semi-analytic and empirical algorithms. The more accurate convolution/superposition (CS) codes use pre-calculated Monte Carlo (MC) dose "kernels", partly accounting for tissue density heterogeneities. Already for many years it is appreciated that full MC simulations of the radiotherapy dose delivery process should result in the highest dose calculation accuracy. Since the second half of the nineties, several MC dose engines for radiotherapy treatment planning have been developed. Recently, vendors of clinical treatment planning systems have started to offer MC dose engines. To avoid excessively long calculation times, approximations and simplifications have been introduced in the algorithms, possibly jeopardising the advantages of full MC dose calculations.

In September 2003 a new NCS sub-commission, focussing on Monte Carlo treatment planning (MCTP) was formed. At that time most vendors of treatment planning software (TPS) were announcing the introduction of MC into their systems, and Nomos had already introduced the Peregrine MCTP dose engine for photon therapy [1]. Therefore a report, providing an overview of the MCTP literature, and an insight into the different techniques used in a MCTP system, was urgently needed. A first draft of this report was finished early 2006, while the report is available since September 2006 as Report 16 of the NCS [2]. The report consists of three main parts:

- Introduction to Monte Carlo
- Fundamentals of Monte Carlo
- Monte Carlo treatment planning in practice

The first part provides an introduction into general Monte Carlo techniques, focussing on applications in radiation dosimetry. The technique is introduced in an accessible way and is

compared to analytical and numerical approaches. The main ingredients of a MC code specific for dosimetry applications, are summarized:

- physics models
- interaction data
- random numbers
- geometry definition
- material composition
- source definition
- scoring
- variance reduction and approximations.

An overview of general purpose Monte Carlo codes (applicable in radiotherapy) is provided. The last chapter of part 1 focuses on the rationale for introducing Monte Carlo into the clinic. A literature overview on comparisons between Monte Carlo results, and dose distributions obtained with conventional TP dose engines, leads to the conclusion that Monte Carlo has certainly proven to provide an added value (even when compared to convolution/superposition algorithms) when studying extreme situations (e.g. a small beam segment impinging on a large air cavity, determining the rebuild-up dose distribution behind that cavity). The extrapolation of these findings to clinical settings is not straightforward though. Therefore published patient studies are summarized as well, leading to the conclusion that most conventional algorithms do not provide the required accuracy. More systematic comparisons between well tuned convolution/superposition algorithms and Monte Carlo dose engines was (and still is) required though before conclusions on the clinical value of MC can be formulated.

The second part of the report consists of a more detailed description of:

- the photon and electron transport, followed by an extensive discussion on the interaction data tables used in the different codes.
- Geometry definition, more specifically the conversion of CT datasets (Hounsfield units) into a voxelized Monte Carlo geometry.
- Modelling of the linear accelerator head, the usage of virtual source models and approximations in the transport through the collimator system (MLC, jaws).
- The scoring, using dedicated scoring grids, focussing on spatial resolution
- Dose to medium versus dose to water
- Variance reduction techniques, denoising
- Inverse optimization
- 4D dose calculations

This chapter should provide medical physicists a clear overview of the main ingredients of a MCTP system.

Part III explains the more practical aspects of MCTP:

- A literature overview of the pioneering work, followed by a description of the most important dose engines developed specifically for MCTP.
- An overview of the commercial TPS vendors that have incorporated (or are planning to incorporate) MC into their software.
- Commissioning: specifically related to the MC option of the TPS. The commissioning process is not that different from that of a conventional treatment planning system. A number of additional aspects should be verified: more specifically the influence of approximations and variance reduction techniques, the influence of noise and denoising, the accuracy of the beam model and the CT conversion. A literature overview on the commissioning of MC dose engines is provided.

The report also contains a number of recommendations. We hope we have succeeded in preparing a document that provides all necessary background information, to allow a medical physicist to select the best available MCTP software package, to commission this system, and to use it optimally in clinical practice. Calculation time is no longer an issue. Most commercial MCTP systems that are currently available are comparable in speed to convolution/superposition algorithms and run on a single PC. It is not clear yet to what extent this speed increase, obtained by using variance reduction, smart programming techniques, and approximations, impacts the potentially very high accuracy of MC systems. Careful comparisons with full-blown MC systems and measurements are needed to assess the quality of the commercial implementations. A shorter version of this NCS report has been published as a review paper in the Radiation Physics and Chemistry journal [3].

After finishing the report, the group decided (in a slightly changed formation) to continue its activities. It was decided to perform extra research to try to determine the added value of Monte Carlo compared to convolution/superposition algorithms. Since 2002, the Belgian lab for standard dosimetry (Ghent University) has developed a Monte Carlo dose engine (MCDE) specifically for post-treatment verification of IMRT patients [4]. The NCS group has planned to use this dose engine to verify several TPS systems used in different centres in Belgium and in The Netherlands. This has already lead to several publications: two convolution/superposition algorithms (Pinnacle and Helax TMS) have been compared with MCDE, both for a group of head-and-neck cancer patients [5], and for lung cancer patients [6]. In the same framework MCDE was used to benchmark Varian AAA for a number of clinical plans for UCL, Brussels [7]. An on-going collaboration with the Erasmus MC hospital in Rotterdam lead to a comparison

between MCDE and the XIO convolution/superposition algorithm (CMS) [8]. Next to a comparison with convolution/superposition algorithms, some commercial MCTP systems will be included as well, when they become available. Additional centres will be included in this study. The setup of this study is illustrated in Figure 1.

This study should eventually lead to a second report of the NCS sub-committee on MCTP, by combining the results obtained in the different centres, possibly allowing a more robust conclusion concerning the added value of MC in the clinic and of a full MC dose engine as QA tool in a standard lab.

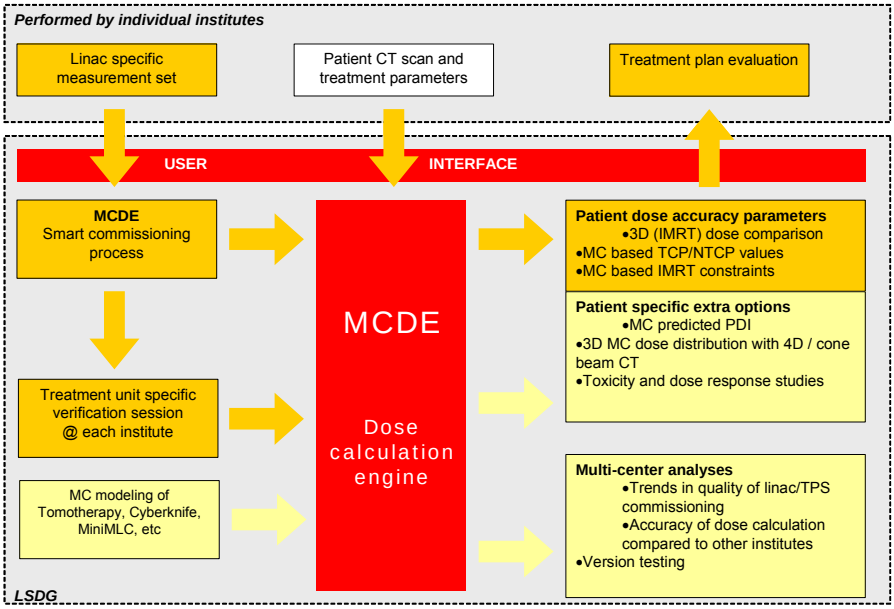


*Figure 1: Treatment plan verification in the Lab for standard dosimetry Ghent (LSDG). A linac head model for the specific centre is tuned starting from a set of measurements provided by the hospital. Once the model is commissioned a verification session, using a dedicated phantom is performed. The centre then submits a plan (in DICOM format) and MCDE is used to determine a 3D dose distribution, that is compared with the conventional planning system. As is illustrated in the figure, one can go very far in this, including portal imaging prediction, the usage of cone beam CT information, and toxicity studies to evaluate the impact of calculation errors in the conventional planning system.*

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# Reference values in diagnostic radiology

J.G. van Unnik and P. Stoop

NCS Advisory platform: Radiology and Nuclear Medicine

The understanding that X-rays are bad for your health is nearly as old as the discovery of the rays. Nevertheless, radiation protection has primarily been aimed at radiological workers in hospitals and industry. The deleterious effects of radiation in patients are felt to be offset by its beneficial effects. Indeed, in most instances, the importance of reaching a diagnosis is far greater than the remote possibility of the patient developing cancer in the future. During the second half of the last century, enormous progress has been made towards reducing radiological exposure to staff and patients.
However, it is not so easy to measure the amount of radiation a patient receives in the hospital setting. As every examination is different in the way the body is irradiated, it is difficult to compare exposure during different examinations and in patients of different body habitus. Various approaches are possible.

One can compare the surface dose in highly standardized examinations as a routine X-ray of the thorax or of the pelvis. This approach has been taken by Shrimpton e.a. in the eighties in their publication of doses to patients from routine diagnostic x-ray examinations in England [1]. They compared the dose of X-ray examinations of the thorax, the abdomen and the limbs in 20 hospitals throughout the UK. Between hospitals the dose was found to vary from 1 to 10. This suggested the possibility of dose reduction. If the radiological dose is established in this way, it is not possible to compare the dose of different radiological examinations with each other or to calculate a collective dose.

Phantom studies can be performed in hospitals with technique settings used for patient studies. The results of these studies can be used to calculate skin entrance dose or effective dose in standard patients. This approach has been taken by Van den Berg et al. in The Netherlands [2] and Mol in Belgium [3].

Alternatively, one can try to calculate the effective dose of a radiological examination from X-ray parameters. The effective dose as defined by ICRP-60 is the sum of the doses to the organs in the body weighted for the risk of cancer induction in each organ. The unity of the effective dose

is the Sievert, but it should be clear that the effective dose can never be directly measured. The calculation of the effective dose and the difficulties of establishing reliable organ doses and weighting factors bring with them a significant uncertainty in the value of the effective dose. This approach was taken by Van Unnik et al. in a study of CT examinations in 14 hospitals in The Netherlands in the 1990's [4]. Again, a 1-to-5 relationship was found between the highest and lowest dose-examinations.

During the past ten years, many publications have established the mean effective dose for a number of examinations [5-11]. In the United Kingdom the National Radiological Protection Board, now merged with the Health Protection Agency, has collected an elaborate database of reference values that has now several times been updated [12]. Spurred by this success the European Commission recommended in 1999 the introduction of national reference values in all member states of the European Union [13]. This recommendation has been picked up in The Netherlands in the new Nuclear Energy Act of 2000. Article BSK 59 states

"Our minister favours the establishment and the use of diagnostic reference values for diagnostic procedures…, as well as the enforcement of specific protocols."

In 1999 the Netherlands Commission for Radiation Dosimetry has created the Platform Diagnostic Radiology and Nuclear Medicine. The platform consists of representatives of the Netherlands Society for Medical Physics, the Dutch Society for Nuclear Medicine, the Radiological Society of the Netherlands, the Dutch Society for Medical Imaging and Radiotherapy and the Netherlands Society for Radiological Protection. The mission of the Platform is:

"The promotion, coordination and support of research on the radiation protection of patients and professionals in radiology and nuclear medicine and the publication of the results of this research."

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# IPEM recommendations on small-field dosimetry: a forthcoming report

H. Palmans
on behalf of the Small Field Dosimetry Working Party of the Institute of Physics and Engineering in Medicine (IPEM): K. E. Rosser, M. M. Aspradakis, J. P. Byrne, J. Conway, S. Duane, H. Palmans and J. Warrington, United Kingdom

## Introduction

Radiotherapy treatment using linear accelerator (linac) generated small photon fields has become an increasing practice in recent years. In stereotactic radiotherapy the application of small fields has been established since many years. The development of multileaf collimators (MLC) combined with improvements in mechanical accuracy and stability and improved dosimetric control has also led to increased usage of small fields at conventional beams. Nowadays, stereotactic treatments can be delivered with mini- and micro-MLC (sometimes as add-ons) for conventional linacs. In intensity modulated radiotherapy (IMRT) a significant dose contribution can originate from small field segments. For commissioning treatment planning systems (TPSs) or independent monitor unit (MU) check systems small field dosimetric data are usually required. Dosimetric errors in small segments could therefore cause in many ways significant treatment errors. A number of interdependent problems exist in the use of small fields; the normal field size definition breaks down; some of the standard dosimetric quantities are difficult to measure and the planning systems' definitions of data may differ from the local clinic's definition. Some of these problems could cause local differences between measurement, prediction and actual dose of the order of 10% [1].

Since several decades, but especially in the last ten years, many scientific publications have appeared on problems associated with small photon fields using conventional linacs and TPSs. The new report by the IPEM Small Field Dosimetry Working Party will review that work and based on this, come up with recommendations that will be helpful in the clinic. The main aims are to summarise and clarify the physics of small field high-energy photon beams, to find out about current usage/practice through questionnaires to UK departments, to elicit manufacturers advice to customers with regards to recommended measuring devices, to carry out a literature review on small field dosimetry and measurement and based on this give guidance on the measurement of these small field beams.

The report will focus on MLC defined field sizes with at least one dimension between 4 mm and 40 mm. The rationale for this somewhat arbitrary choice is that for field sizes larger than 40 mm conventional dosimetry is assumed to be sufficiently accurate whereas for field sizes smaller than 4 mm the available literature is deemed insufficient to give comprehensive recommendations and advise.

The report is at present work in progress and not all parts have been worked out yet. An introduction to the findings on the main problems in small field dosimetry and the use and availability of detector and Monte Carlo simulations is given here. The aims of those aspects that are less elaborated at this stage such as validation of small field data and comprehensive recommendations will be briefly discussed in the last section on future developments.

**Problems of small-field dosimetry**

The beam parameters that usually need to be measured as basic input for a TPS are lateral profiles and penumbrae, output factors or reference dose levels, depth dose distributions, head scatter and phantom scatter factors. For each of those considerable problems can occur when small fields are used some of which (but not an exhaustive list) will be given below. For a start, one must be aware that a TPS optimised for large fields is unlikely to give good intensity maps when small fields are included in the plan. Even for conventional sized treatment fields TPSs often require small field data in beam modelling. Inaccurate modelling can lead to inaccurate MU prediction and inaccurate positioning of MLC leaves potentially resulting in hot or cold spots in IMRT delivery. It may be necessary to consider separate TPS models for large and small fields if it is found difficult to implement accurately all beam sizes ranging from 4 mm to 400 mm with a single model.

An immediate dosimetry problem occurs when inspecting lateral profiles of small fields as shown in figure 1. It is obvious that below 40 mm the curvature of the central plateau region will start to interplay with the lateral extension of a detector with a diameter of 10 mm (a typical ionisation chamber size used in the measurement of large field output factors). One has thus to rely either on smaller ionization chambers, with loss of sensitivity and more potential problems with leakage currents, or on alternative detectors such as diodes, diamonds or film, with additional uncertainties due to the lack of linearity of the dose and dose rate response or due to an energy dependence of the response.

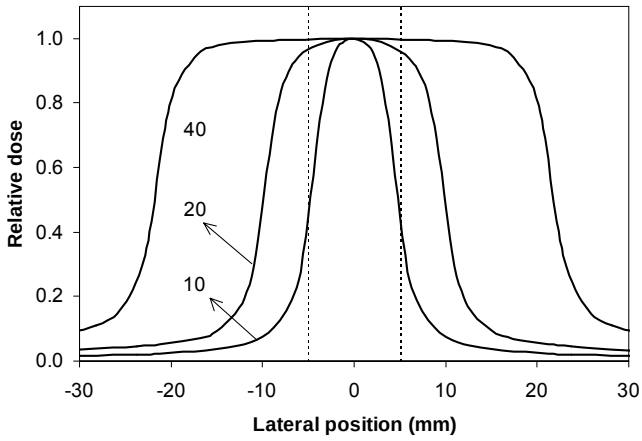


*Figure 1: Lateral profiles for slit fields of 100 mm long and 40 mm, 20 mm and 10 mm wide measured with a diamond detector by De Vlamynck et al (1999) [2]. The dotted vertical lines indicate the size of a 10 mm detector.*

Two important dosimetric effects in small fields are related to shielding of the secondary photon sources. When the MLC and jaws close they will gradually shield more of the scattered photons generated mainly in the primary collimator and the flattening filter. The first effect this has is that the penumbrae will get narrower since the blurring effect on the penumbra of the extended secondary photon sources will be reduced or largely disappear. A second effect is that the output factor will be reduced because of the reduced number of (secondary) photons exiting the final collimator system. The reduction in output factor is further enhanced at the smallest field sizes when the penumbrae start to overlap. This is in part the consequence of the loss of lateral charged particle equilibrium. This explains the steeper decrease of the output factor as a function of field size at the smallest fields as is visible in figure 2.

The decrease in output factor with decreasing field size, especially there were it drops steeply due to the overlapping penumbra also causes an apparent widening of the beam with respect to the collimator setting when the full width at half maximum (FWHM) of the beam profile is used as the measure of field size. This observation questions the unambiguous definition of field size. Even though this definition may be useable depending on the requirements of the TPS, an alternative definition could simply be the collimator setting provided each collimator's zero position is on the collimator rotation axis. The main message is that the way field size is measured should be consistent with the definition assumed by the TPS.

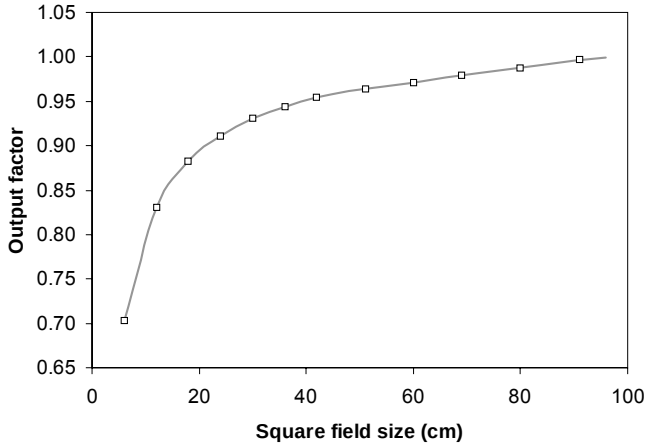


*Figure 2: Output factor at 50 mm depth as a function of square field size measured with a photon diode in a Varian CL2300 6MV photon beam mounted with a Brainlab micro multileaf collimator (data digitised from the paper by Belec et al 2005 [3]).*

The size of the primary photon source is an example of a parameter that may not be critical in modelling large fields but is very critical for small fields. Another example is the modelling of the MLC where commonly made approximations that have little impact for large fields affect accurate dose prediction in small fields; Lydon (2005) [4] found that a TPS did not model the rounded leaf ends of the Varian units and though it was an insignificant approximation for larger field sizes, it introduced significant errors in small field modelling.

**Detectors**

The problem of detector size has been mentioned above. Traditionally, ionisation chambers with volumes roughly between 0.1 cc and 1 cc have been the basis of reference dosimetry in radiotherapy since they exhibit a good sensitivity to radiotherapy level doses and are small compared to conventional field sizes. However, they can become unsuitable in the presence of high dose gradients but also in the presence of time-dependent dose variations and varying energy fluence. In addition, volume averaging and lack of electronic equilibrium complicate the use of ion chambers for the dosimetry of small photon beams. As a response to this problem, micro-chambers with volumes in the order of 0.01 cc and smaller have been introduced (e.g. Martens et al 2000 [5]), which have the advantage of a more suitable size for field dimensions down to 20 mm but, on the other hand, have a low sensitivity and often exhibit a more pronounced energy dependence to low-energy electrons. These small volume ionization chambers are therefore complemented by a range of other dosimeters such as film, diode and diamond detectors, each having certain dosimetric advantages.

A number of criteria to describe the ideal detector could be brought forward: it should return absolute dose to water as output, it should have a stable response in time, its response should be linear as a function of dose, dose rate and dose per pulse, it should be water equivalent both in the sense of energy independence as in the sense of unaffected by perturbations, its response should be orientation independent, it should be small compared to the geometrical extension of the measured dose variations, it should exhibit no significant background reading, its response should be independent of environmental conditions as temperature, pressure and humidity (or at least an accurate model to correct for these influence factors should be available) and it should have an appropriate time resolution for the measurement of time dependent output quantities. Obviously however, no single detector will incorporate all these ideal characteristics. It is therefore more sensible to approach the selection of the most suitable detector from the requirements for a specific measured quantity.

For example for the measurement of penumbra and beam width, spatial resolution must be small [1, 6] but in addition it should respond linearly to dose and dose rate given the large difference between dose in and outside the field. It should also be insensitive to the spectral differences inside and outside the field. Some solid-state detectors, although ideal from the perspective of resolutions, may have problems with the other criteria. On the other hand, it can be possible to determine accurate correction factors for those non-ideal properties. Also a detector with a coarser resolution should not necessary be excluded. It is well known that the larger the detector size is the greater the measured penumbra will be [7] but methods have been developed to derive the true penumbra from a detector measurement using a detector specific kernel obtained by comparing the detector with film in a step edge profile, assuming film gives the correct penumbra [1].

Another interesting example is the measurement of output factors, for which both in MLC defined [1] or stereotactic radiosurgery beams [8] large differences between detector types have been demonstrated. Given the need for measuring absolute dose at the dose maximum, resolution is again important but the measurement of output factors in the smallest fields is also characterised by the lack of electron equilibrium potentially resulting in significant detector perturbations. From this point of view water equivalence is another important property of the detector and hence, a diamond detector would be preferable over a diode.

For the measurement of depth dose distribution, tissue phantom ratios, etc. the requirements for a suitable detector are more relaxed although given the change in the beam spectra with depth the lack of water equivalence of some detectors such as diodes may again result in an under or over response [6].

**Monte Carlo simulations**

Monte Carlo simulations are found to be an invaluable tool supplementing measurement and enhancing understanding of small field dosimetry given its ability to track individual particles, to calculate local energy spectra, to score different quantities in a single simulation (e.g. kerma and dose), to separate different components contributing to the local dose (e.g. from primary and scattered photons), to calculate dose in an arbitrary small volume and to incorporate the detector geometry in the calculation model. Another strength of Monte Carlo is that in theory, no approximations about the beam model need to be made even though in practice, given the limited accuracy with which parameters like the energy and the lateral extension of the primary electron beam are known, some tuning of beam parameters is required.

Implementing Monte Carlo simulations for small field dosimetry requires three steps; an accurate beam model has to be built, the system has to be tuned and commissioned by validation with measured data and the small field problem has to be set up often including a detailed description of the detector geometry. Regarding beam modelling of high-energy clinical photon beams an extensive overview (including small field) is given by Verhaegen and Seuntjens (2002) [9]. The main parameters that require tuning are the mean energy of the incident electron beam, usually based on matching the simulated and measured depth dose distributions, and the lateral distribution of the beam spot (assumed to be gaussian) usually based on matching the simulated and measured lateral profiles. Subsequent small field simulations require care with respect to the geometrical resolution of the dose scoring, selection of appropriate transport parameters and the awareness that it might be necessary to include a full geometrical model of the detector in order to understand its behaviour.

Some particular examples of cases were Monte Carlo simulations have proven to be very useful are the calculation of scatter factors [10], component analysis of output factors [11], study of lateral electron equilibrium [12], extending measured data to smaller field sizes [13], the simulation of ionization chamber perturbation correction factors [14, 15, 16] and dose perturbations in inhomogeneities [17, 18].

**Future developments**

Based on the literature review, recommendations will be formulated on the aspects that require special attention and caution, the quantities to be measured, the choice of suitable detectors for particular measurements, the implementation of Monte Carlo simulations and the validation of small field dosimetric data. An overview of commercially available detectors and their characteristics will also be given as well as sample data obtained according to good practice procedures. It is expected that the report will be published in 2008.

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# Revisiting film dosimetry in radiation therapy

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Film “dosimetry” is closely connected to the discovery of X-rays by Roentgen in December 1895, so that film can be considered as the oldest radiation dosimeter. Radiographic films based on silver halide emulsions are still widely used for 2D dosimetry in radiotherapy. In the current IMRT epoch, film dosimetry has become even more indispensable for verifying complex dose distributions. Palm *et al.* expressed it as “Radiographic film dosimetry is experiencing a renaissance in the radiation therapy community” [1]. Film has always been an intuitively clear and conceptually simple integrating dosimeter that offers a unique planar spatial resolution. In the last decade, the availability of affordable scanning systems allowed film dosimetry to enter the digital arena and to thoroughly assess its accuracy and to reveal some weaknesses.

In this contribution, we will not reiterate material that is covered in the still growing large body of literature on film dosimetry. For a literature overview, we refer the reader to the recent report of AAPM’s task group 69 [2]. Instead, this presentation will give a practical overview and will touch some fundamental peculiarities and challenging riddles that have not been completely clarified yet.

Radiographic film is known to have a photon energy dependent response. Because of the high atomic number of silver, photoelectric interactions in film become important for photon energies below 200 keV [3, 4]. Consequently, film sensitivity increases with field size and depth due to an increasing contribution of low-energy Compton scattered photons. However, there is no consensus about the extent of the deviations. For instance, Burch *et al*. [5] found that, for a 4-MV photon beam evaluated at 5-cm depth, the response increases with ~5% when increasing the field size from 6×6 $cm^2$ to 25×25 $cm^2$, while Sykes *et al*. [6] did not observe any effect of field size for identical irradiation conditions and the nominally same 4-MV beam quality. When analysing depth dose curves, Burch *et al*. [5] found for the 25×25 $cm^2$ field an increase in sensitivity of 12% when increasing the measuring depth from 5 to 15 cm. According to Sykes *et al*. [6], however, this sensitivity increase was only 6%. The most common types of radiographic film are Kodak XV2 and EDR2 (only available since 2002). Relative to XV2 film, EDR2 film has been reported to have less dependence on the processor, depth and field size [7, 8]. A lower photon energy dependence for the EDR2 film was reported and attributed to the reduced silver

content and smaller grain size [7]. Because of its enhanced dynamic range, EDR2 film can be used to measure a complete fraction of an IMRT delivery.

The sensitometric curve of film plots optical density (OD) as a function of dose to the medium, principally in the absence of the film [9]. This sensitometric curve, sometimes called H&D curve after Hurter and Driffield, depends also on the photon or electron beam energy, film emulsion, development process and scanner. Some groups propose a common "unique sensitometric curve" per centre, which may be sufficient for relative dosimetry if processing conditions are well controlled. Van Battum and Huizenga [10] followed a single-hit model for XV2 film while Georg *et al.* [11] normalized 6-MV and 25-MV curves for XV and EDR2 film using a two-hit model. For accurate film dosimetry, however, most researchers determine a sensitometric curve for each experiment and process the calibration films together with the experimental films. The calibration conditions should further match the experimental ones as close as possible. Ideally, depth, field size, photon beam quality or electron beam energy, relative orientation to the beam and dose rate should be identical. In addition, film packaging during calibration should be the same as during the actual dosimetry because of the possible extra film blackening due to Cerenkov radiation [12]. We found that removing the white paper from the Kodak Ready Pack may lead to a dose response decrease of 7.9% [13] [T. Vercauteren en C. De Wagter, unpublished data, June 2006]. Figure 1 displays typical sensitometric curves for XV2 and EDR2 film. The linear (XV2) and parabolic (EDR2) onsets are characteristic for the single- and double hit process respectively. It is clear that, for clinical treatment verification, EDR2 film is to be preferred over XV2 film because EDR2 film can handle a dose of at least 2 Gy without saturation, albeit at the expense of dose resolution. Interestingly, the Kodak Ready Pack is waterproof allowing to perform radiographic-film dosimetry in watertanks [14].

For dosimetry of individual IMRT beam segments, when the film is oriented perpendicular to the beam axis at a depth of typically 5 or 10 cm, film dosimetry is considered reliable in both the high-dose and low-dose parts of the field. Martens *et al*. [15] found that for equivalent field sizes up to $15\times15$ cm$^2$, the deviations remained within 3% for XV2 film at 6 MV and 18 MV. Yeo *et al.* [16], on the contrary, obtained higher deviations at 6 MV for the same measuring conditions. They could, however, reduce the overresponse in and outside penumbra regions from 9% to 3% by using thin lead foils parallel to the EDR2 film, extending earlier work on XV2 film by Burch *et al*. [5]. Interestingly, Palm *et al.* [17] observed that XV2-film overresponse also increases with phantom size, viz., with the lateral scatter material. For a $20\times20$ cm$^2$ field at 20 cm depth the film overresponse measured in a 50 cm square polystyrene phantom was twice that measured in a

25 cm square phantom, 16% and 8%, respectively. We were not able to reproduce these results [unpublished material].

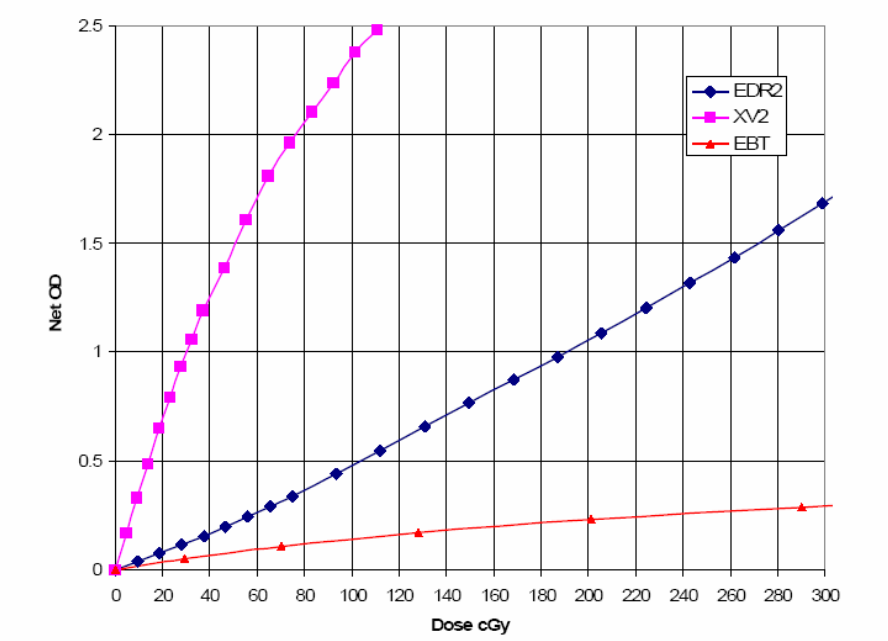


*Figure 1: Sensitometric curves for radiographic film (EDR2 and XV2) and radiochromic film (EBT)*

Although radiographic film is widely used as "composite film" dosimeter for entire-treatment dose verification in IMRT, its validity is lower then and even more conflicting data have been reported. The crucial point is that the film response is depth and beam orientation dependent, and that the two may vary from beam segment to beam segment. Both XV2 film [18, 19] and EDR2 film [20] show a higher sensitivity in the region around dose maximum, typically by 4%, when the film is oriented perpendicular rather than parallel to the incoming radiation. In Figure 2, we plotted the corresponding sensitometric curves for both orientations of a 6 MV beam. This chart however is in total contradiction to that published in [8]. Removing the air layers along the film surface by exerting enough mechanical pressure after puncturing the package is mandatory in parallel film dosimetry [20]. The difference of 4% resulting from Figure 2, however, is not due to perturbation effects by the upstream parts of the film as assumed in [21]. In Figure 3, we obtained the same 4% difference for small pieces of film and proved hence that the orientation dependence is an intrinsic and local effect.

In an inter-centre QA intercomparison of IMRT verification, the European QUASIMODO group used pelvic phantoms that contained seven EDR2 films in transverse planes of the phantom [20]. The original intent was to interpret the "composite film" dose values in an absolute way, i.e. without normalisation using ionisation chamber measurements. However, presumably due to the

earlier mentioned problems with composite film dosimetry, the authors had to resort to a 2-parameter linear conversion from the “film dose” to the actual dose to water in order to fit the film data to point measurements that were taken in both a high- and low-dose region.

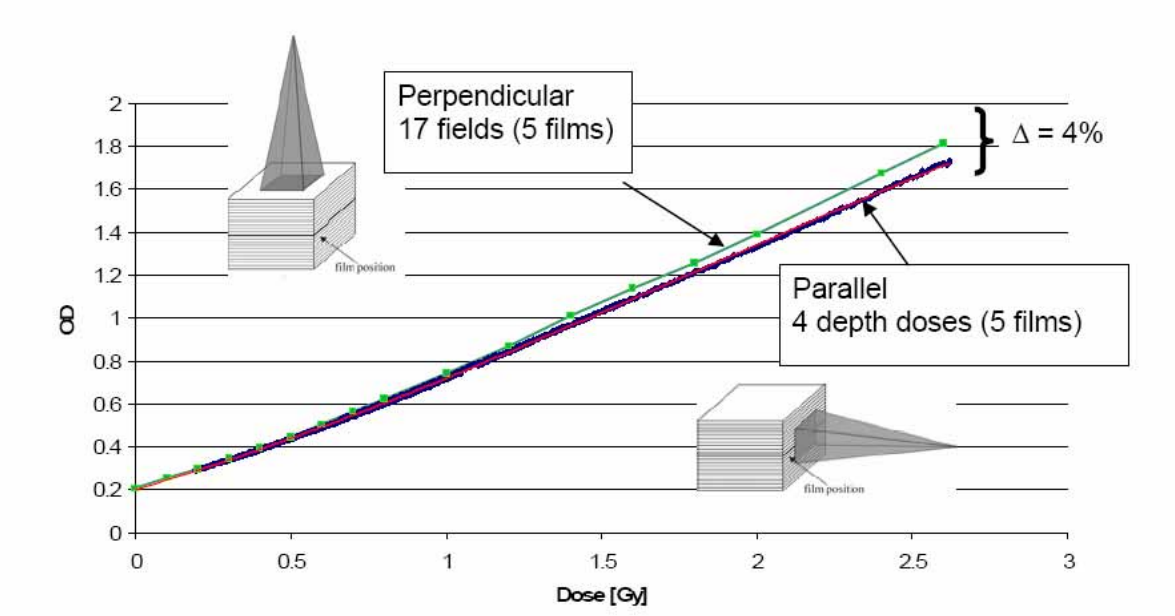


*Figure 2: Effect of film orientation on the sensitometric curves for EDR2*

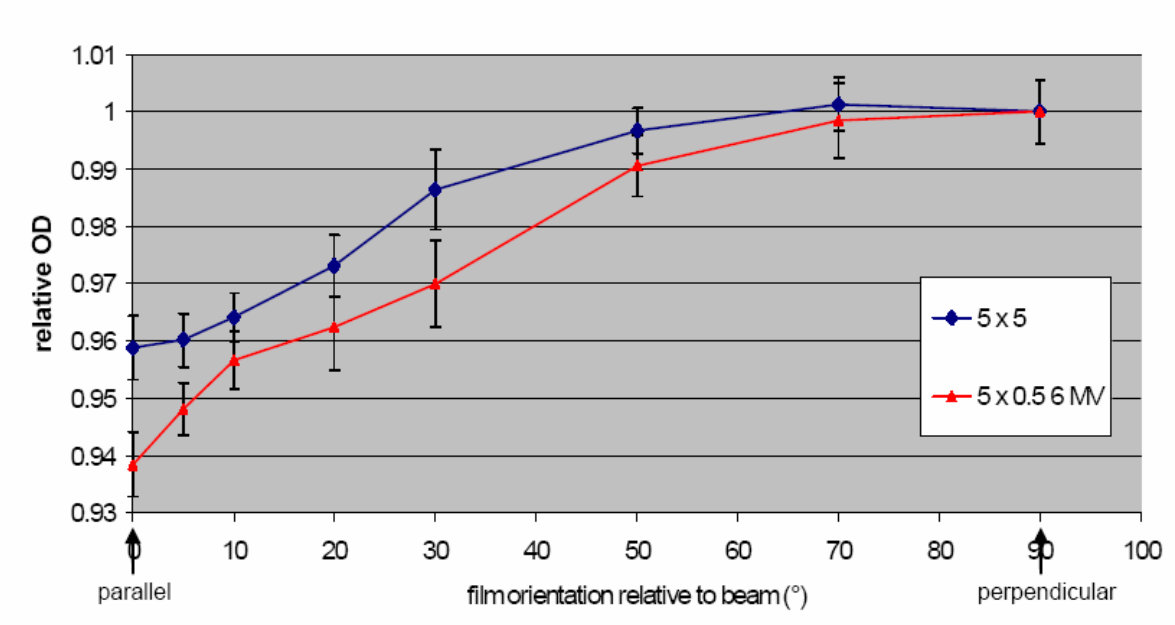


*Figure 3: Intrinsic directional dependence of small pieces of EDR2 film for ($5 \times 5 cm^2$) and ($5 \times 0.5 cm^2$) fields at 6 MV*

In a recent publication [22] we demonstrated that EDR2 film is also a valuable dosimeter for compensator-based IMRT. A film dose underresponse of 1.1% was the maximum error found, which occurred for a 30-mm thick MCP-96 block in a 25 MV beam, which realized a transmission factor of 0.243. The effect induced by the compensators is higher than the experimental error but still within the accepted overall uncertainty of film dosimetry in clinical IMRT QA. These results contradict data from Wiezorek *et al*. [23] who claimed that EDR2 film features a 5% underresponse for a 32-mm thick block.

As any dosimeter, radiographic film is basically affected by the electrons [24] – rather than by the photons – and can therefore be considered as a reliable detector in electron beams [25]. However, contrary to the Kodak XV film, EDR2 films exhibit an energy-dependent sensitivity enhancement for electron beams relative to photon beams [26].

The accuracy of film dosimetry and the useful dynamic range are determined by the film type, developing process and scanner. An interesting and quite general treatise on film scanners can be found in [2], inclusive the preferable digital file format. A general characteristic of these scanners is that they introduce noise near the high end of the useful dose range, because of the lower signal-to-noise ratio at high ODs [27, 28].

Radiochromic films, on the other hand, do not contain heavy elements like silver and are practically water-equivalent. As they are irradiated, they become bluish due to the polymerization of a radiochromic dye. By virtue of their tissue-equivalence [29], there are no concerns about energy dependence and perturbation effects. The first radiochromic films however were small-sized and expensive. Their dose response was inhomogeneous and was affected by post-irradiation colouration. The dose sensitivity was low and their application was labour-intensive. Since the introduction of Gafchromic EBT film (available since 2004), which is almost free of previous drawbacks, radiochromic film has become an important player in film dosimetry. The limited OD range, as apparent from Figure 1, and the resulting low dose resolution impose high demands on the optical scanner [30]. As shown in Figure 4, radiochromic film can be accurately trimmed in shape to the local contours of the phantom. Radiochromic film allows a direct insertion in water phantoms, although with minor artefacts of (visible) water penetration at the film edges [31].

With regard to the quality assurance of film dosimetry itself, we should realize that the film is a mere link in the chain that comprises the film, phantom, developer and scanner. Each component of the chain requires its own quality control.

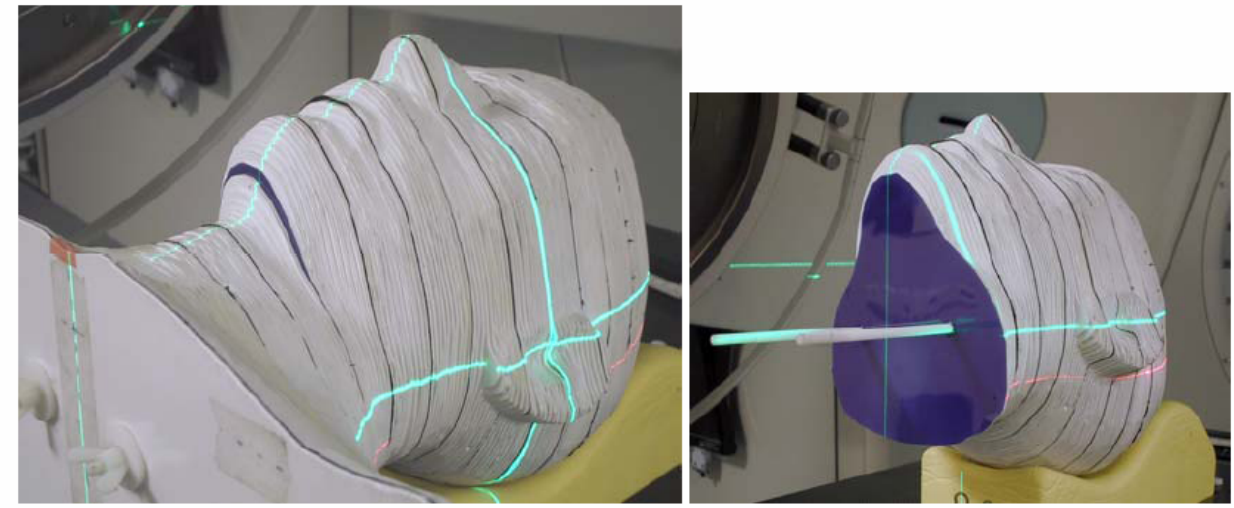

*Figure 4: Head-and-neck phantom transversely loaded with radiochromic EBT film for IMRT treatment verification*

In conclusion, revisiting film dosimetry in the context of IMRT is a worthwhile exercise. Film is and will remain a well-established 2D dosimeter with an unquestioned spatial resolution and intuitive ease to use. With the impetus of radiochromic film, we may further expect an improvement of accuracy and applicability.

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## Radiation exposure as a result of PET/CT

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**Summary**

This prospective study in the field of $^{18}$F-FDG PET/CT was aimed at

1) quantifying the radiation dose to workers, patients and hospital environment,
2) optimisation of procedures,
3) education of workers.

TLD100 (LiF) was used for ambient and finger dosimetry, electronic pocket dosimeters for general occupational dosimetry. In the optimization of the working protocol a few alternatives were evaluated: two shipping containers with different wall thicknesses for delivery of the activity to the department, two dispenser systems for filling syringes with activity, and either with or without a mobile shield during actions related to venous access to the patient. All worker related results are presented for a single nuclear medicine technologist (NMT) performing 1000 patient studies per year, each patient receiving an activity of 370 MBq $^{18}$F-FDG.

The results of the ambient dosimetry confirmed the status of the 1, 6 and 20 mSv/y zones.

The thicker shipping container for FDG-delivery significantly lowered the dose to the NMT. When $^{18}$F-FDG was dispensed using the Koenders system, the total finger dose was on average 8.4 mGy; for the second dispenser, the Docking station, this value was 71 mGy (p=0.000). Injection of $^{18}$F-FDG, using the Koenders system, caused a finger dose of 4.6 mGy; the second system resulted in 23 mGy (p=0.000). Using the mobile shield during injection of FDG, connecting the I-contrast for CT and removal of the intravenous access, also lowered the NMT dose significantly. Combining the alternatives causing the lowest occupational exposure, i.e. the thick container for delivery of FDG, the Koenders dispenser system and use of the mobile shield, resulted in a total effective dose of 2.8 mSv for the NMT.

The total effective dose for a patient undergoing a PET/CT of the head of 2 or 3 bed positions was 10 or 11.6 mSv, respectively, undergoing a whole body PET/CT of 7 or 8 bed positions caused 19 or 20 mSv, respectively.

## Introduction

After essentially having been a research instrument for at least two decades, PET is now seen as an indispensable tool in oncological imaging. The PET-tracer that is most widely used is $^{18}$F-2-fluoro-2-deoxy-D-glucose ($^{18}$F-FDG ), a glucose analogue that in many tumours is taken up at a higher rate than in normal tissue. This high uptake makes it possible to detect primary tumours and metastases that might have been missed in images from the common radiological modalities CT, MRI and ultrasound. The present state of the art PET-scanner is a hybrid system that includes a CT-scanner, providing co-registered images of the two modalities. The CT-images are commonly used for attenuation correction of the PET-data and for anatomical reference. The latter adds significantly to the overall performance of a PET-study that generally lacks adequate anatomical information.

$^{18}$F has a half-life of 109.8 min, a $\beta^+$-abundance of 96.7 %, and a maximum energy of the beta-particle of 0.634 MeV. According to ICRP 53 [1], after injection $^{18}$F-FDG is cleared from the circulation with a halftime of less than 1 minute. The uptake in the brain and the myocardium is 6 % and 8 % of the injected activity respectively, with an uptake halftime of 8 min. About 30 % of $^{18}$F is excreted in the urine. The initial metabolism of $^{18}$F-FDG is identical to that of glucose, but then it is not a substrate for further metabolism, leaving the $^{18}$F where the FDG was taken up. Total body retention can be characterized with 3 fractions of 7.5 %, 22.5 % and 70 % of the injected activity, with half-lifes of 12 min, 1.5 h and infinity. The large fractions with the long half-life facilitate imaging over a period that is essentially limited by the physical decay of $^{18}$F. After $^{18}$F administration PET-scanning is postponed for between 45 min and 2 h, because it is found that the contrast between tumour en normal tissue increases with time, outweighing physical decay. The effective dose to the patient is 0.019 mSv/MBq; the bladder wall is the organ with the highest absorbed dose, 0.016 mGy/MBq [2].

Studying the occupational radiation exposure caused by PET imaging is warranted because each positron emission is associated with two 511 keV gamma photons that are far more energetic and penetrating than e.g. the 140 keV radiation of $^{99m}$Tc, the “work horse” of non-PET nuclear medicine. The differences are reflected in the kerma-in-air-rate-constants ($\Gamma$’s) and the half-value layers: the ratio $\Gamma(^{18}\text{F})/\Gamma(^{99m}\text{Tc})$ is about 7.5, the half-value layer in lead is 3.9 mm for $^{18}$F versus 2.8 mm for $^{99m}$Tc, and in water 6.9 cm versus 4.3 cm. Moreover, in the literature relatively high exposures of workers have been reported [3-8], although several studies also found more acceptable values [9-11].

Rather than reviewing the literature, a study is presented that might be useful to perform in any department using PET/CT. This prospective study was executed in the azM and aimed at 1) quantifying the radiation dose to workers, patients and hospital environment, 2) optimization of procedures, and 3) education of workers.
In this contribution only part of our work is presented. For more information see [12] .

**Materials and methods**

The PET/CT-scanner available was a Siemens Biograph Sensation16. All dosimetry was performed under normal working conditions. The number of patients was registered, as well as the total $^{18}$F-activity that was handled and the activity that was administered to the patients.
The radiation exposure was quantified for nuclear medicine technicians (NMTs), radiotherapy technicians, patients, and indirectly, by measuring ambient doses, other hospital staff and members of the public. We used two types of electronic personal dosimeters (EPDs) for occupational dosimetry, the Mini Instruments 6100 Series Dosimeters (Saint-Gobain Crystal & Detectors UK Ltd), and the MK2 (Siemens Environmental Systems Ltd, Dorset, UK). In addition we used TLD100 thermoluminescence dosimeters for finger and ambient dosimetry (TLD 100: LiF, 0.9x3x3mm$^3$, Harshaw, Stratec Services BV, Houten, the Netherlands). The Mini Instrument measured Hp(10), the MK2 both Hp(10) and Hp(0.07), while TLD100 was calibrated in air-kerma. The EPDs were connected to the chest pocket of the NMT's dress; we will refer to this position as "chest". For ambient dosimetry TLDs were positioned at many locations and left in position for several months.
Effective dose (E) to the workers was approximated with the average of the measured Hp(10) on the chest and the estimated exit dose assuming a worker is three half-value layers (21 cm) thick, i.e. E = (9/16)*Hp(10).

After being injected with $^{18}$F-FDG the patient rested for 45 minutes in a shielded room with remote surveillance. After visiting the toilet, the patient was positioned on the scanner bed. First the CT scan was performed, often using I-contrast, followed by the PET scan. Patient dose due to the PET study was estimated using the ICRP 80 effective dose of 0.019 mSv/MBq $^{18}$F-FDG. For the calculation of the CT dose the Impact CT Patient Dosimetry Calculator was used [13].

Alternative procedures were compared:
1) using lead containers, holding the $^{18}$F-FDG upon delivery to the hospital, with different wall thicknesses (17 mm and 30 mm).

2) dispensing $^{18}$F-FDG to syringes with the 'Koenders system' ('Shielding device for PET patient doses', von Gahlen, Didam, The Netherlands) and the 'Docking station' ('Protective vial container for 511 keV with Docking system', Veenstra, Joure, The Netherlands).
3) with or without a mobile lead shield during administration of $^{18}$F-FDG and removal of the intravenous line (shield: 20 mm Pb-equivalent, Medisystem, Guyancourt, France).

Dispensing of $^{18}$F activity is performed in a shielded laminar flow cabinet. When the Koenders system is used, the vial with the stock $^{18}$F-FDG remains in the dose calibrator and activity is withdrawn through thin tubing into a syringe in a well shielded container [14]. This container can also be used during injection. The activity in the syringe is given by the difference in reading of the dose calibrator before and after withdrawal of activity. The bulk of activity is always shielded, only the activity in the thin tube is not, but only as long as the line is not flushed with saline. The Docking station consists of a heavily shielded container for the $^{18}$F-FDG vial. Activity is withdrawn into a syringe protected by a conventional syringe shield with a lead-glass window. This is facilitated by placing the syringe with shield in a docking system that is connected to the container with vial. For measurement of the activity, the syringe has to be taken out of the syringe shield.

In the normal routine the following steps were distinguished (underlined are the alternatives studied):

1. Daily quality control using the 6 litre $^{68}$Ge/68Ga barrel phantom (15 – 40 MBq),
2. Unpacking of container (17 or 30 mm wall) with $^{18}$F-FDG and transport of it to hotlab,
3. Filling syringes with $^{18}$F-FDG (using either Koenders system or Docking station),
4. Placing the shielded syringes with $^{18}$F-FDG into a mobile lead container for transport,
5. Transport of this $^{18}$F-FDG to the room where the patient will be injected,
6. Injection of the $^{18}$F-FDG via an already established intravenous line (w or w/o shield),
7. Sending the patient to the toilet after a 45 min wait,
8. Taking the patient to the scanner,
9. Positioning of the patient in the scanner,
10. Connecting the line for iodine contrast for optional diagnostic CT (w or w/o shield),
11. Removal of the intravenous line (w or w/o shield),
12. Patient off table and leaving the department.

Note that during the CT-scan no worker is present in the room. After each step the reading of the EPD was written down. All occupational dosimetry results were normalized to 1000 patient

studies and 370 MBq $^{18}$F-FDG administered to the patient. Thousand patients corresponded approximately to the workload of 1 year in our department.

**Results**

Ambient dosimetry showed that the limits in the 1 mSv/y, 6 mSv/y an 20 mSv/y zones were nowhere exceeded.

Occupational dosimetry was performed during approximately 750 patient studies. There was a large difference in finger dose (air kerma) associated with the use of the two dispenser systems (Figure 1). The Koenders system provided a much better protection of the fingers than the Docking station. A similar large difference was observed while injecting the activity into the patient.

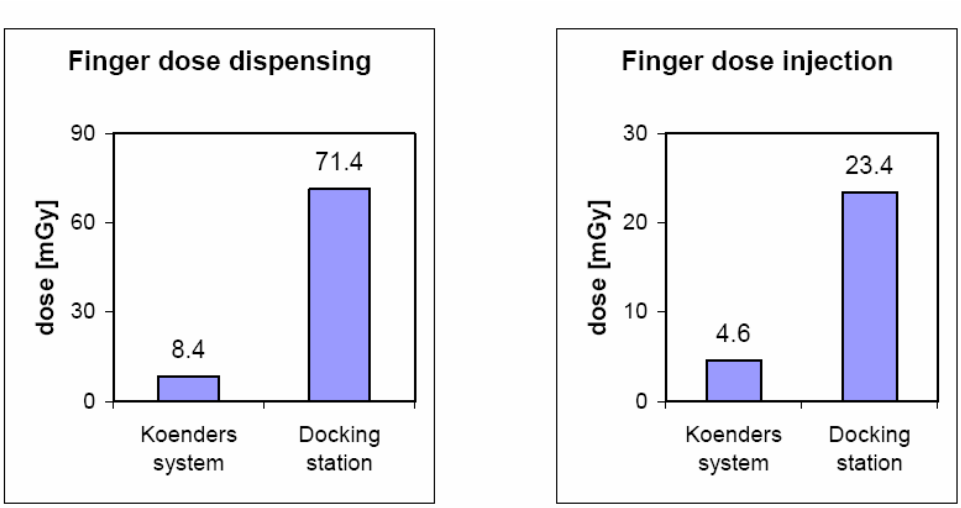


*Figure 1: Nuclear Medicine Technician finger dose (air kerma) incurred during 1000 studies of 370 MBq using two different dispenser systems (Koenders system and Docking station). Left graph: filling syringes. Right graph: patient injection.*

When the syringe with $^{18}$F-FDG is handled without shielding, the skin of the hands is also exposed to beta radiation. Using the MK2 it was found that the $\beta^+$-radiation accounts roughly for 50% of Hp (0.07). Most of these beta's were stopped by the cap of the finger ring that contains the TLD. In the rare case that a technologist holds an unshielded $^{18}$F-FDG filled syringe in his bare hands, the finger doses derived from TLD measurements may have to be doubled.

The difference in Hp(10) at the chest as measured with the EPD for the various alternatives we studied are shown in Figures 2 – 4. Figure 2 shows the advantage of using a 30mm walled lead container for $^{18}$F-FDG delivery to the hospital instead of one with a 17mm thick wall.

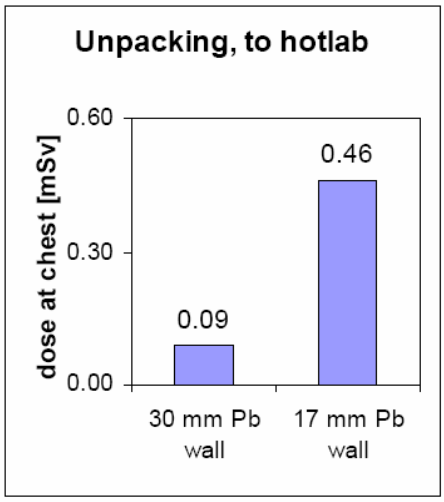


*Figure 2: Dose (Hp(10)) at the chest of the MNT for 2 different shipping containers used for delivery of $^{18}$F-FDG to the hospital for 1000 studies of 370 MBq $^{18}$F-FDG.*

In Figure 3 the chest dose is shown when using the two different dispenser systems for filling the syringes and injecting the patient, the latter with the mobile lead shield in place. Again, the Koenders system helps best to keep the dose to the NMT low.

In Figure 4 the benefit of the lead shield while injecting the activity and removing the intravenous line is illustrated.

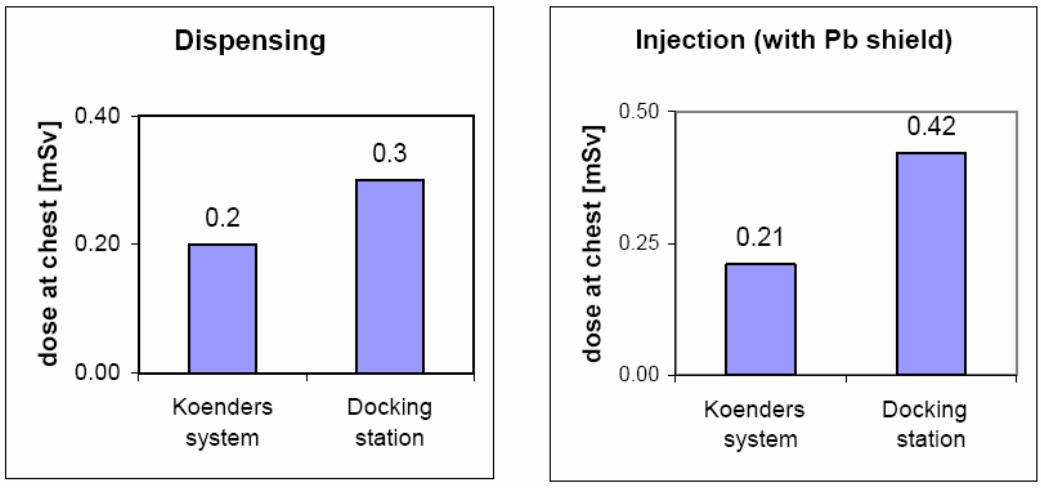


*Figure 3: Dose at the chest of the MNT due to 1000 studies of 370 MBq for two different dispenser systems (Koenders system and Docking Station). Left graph: filling syringes. Right graph: patient injection (with the additional mobile shield in place).*

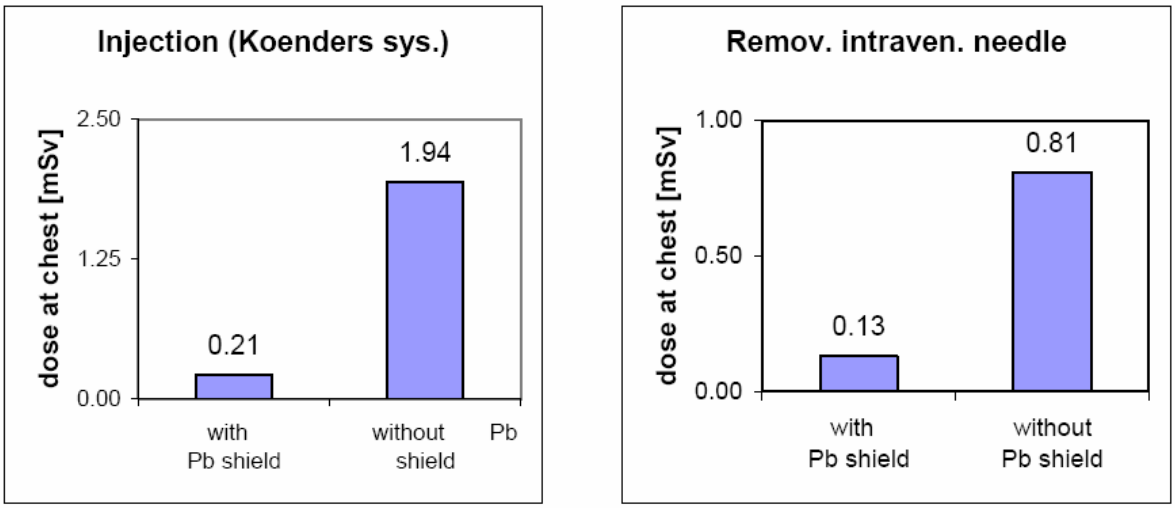


*Figure 4: Dose at the chest of the MNT with and without using the mobile lead shield for 1000 studies of 370 MBq. Left graph: injection of the patient. Right graph: removal of the intravenous access.*

Upon completion of this study the standard procedure was chosen to include the use of the 30 mm walled lead container, the Koenders dispenser/injection system, and the mobile lead shield. The contributions to Hp(10) at the chest from the various steps in the investigation of patients using this protocol are shown in Figure 5. The corresponding effective dose to a NMT performing 1000 patient studies of 370 MBq is estimated as 2.8 mSv (3.9 mSv before optimisation).

Radiotherapy technicians, who only come to our department for positioning patients for radiotherapy planning scans, receive a considerable effective dose when compared to NMTs who are involved in all steps of the studies: 2.0 mSv per 1000 studies of 370 MBq. Although a normal PET/CT procedure is performed, patient positioning requires additional attention because the images have to be used in radiotherapy planning.
The effective dose to the patient due to receiving 370 MBq $^{18}$F-FDG is 7.0 mSv. The diagnostic CT scan causes, in the case of a head scan of two (three) PET-bed positions, an additional effective dose of 3 mSv (4.6 mSv). A whole body study CT results in an effective dose of 12 mSv (13 mSv) corresponding to seven (eight) bed positions. For CT scans that are only to be used for attenuation correction and anatomical reference, a considerable lower load (mAs) of the X-ray tube can be applied, resulting in a decrease in effective dose by a factor of 0.2 to 0.3.

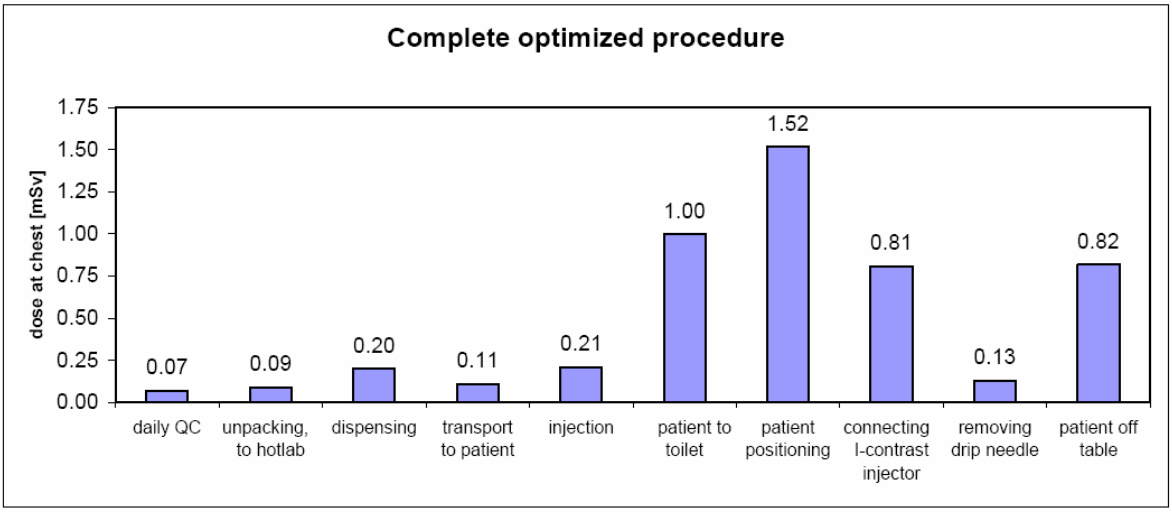


*Figure 5: Dose at the chest of the MNT due to the various tasks in the complete optimized patient investigation (1000 studies of 370 MBq). Note that some steps mentioned in Materials and Methods were combined.*

## Discussion

Quantitative assessment of radiation exposure due to PET/CT requires some effort, but the benefits are considerable. These advantages include the possibility of optimisation of procedures, of comparison of quantitative information with guidelines and legal rules, and of making workers apprehensive about their own influence on their exposure.

It was found that the finger dose can be kept low by carefully shielding the activity. The Koenders system performs very well in this respect. Only when the activity is passing through the thin tubing between stock vial and patient syringe it is unshielded. However, since the plunger of the syringe is moved with long tweezers, the distance between the hand and activity is large. The Docking station uses a syringe in a conventional syringe shield, and although considerably heavier than for $^{99m}$Tc, it provides less protection than the still heavier container of the Koenders system that surrounds the syringe more fully. Moreover, for the Docking station the syringe has to be taken out for measurement of the activity.

With the shielding of $^{18}$F-FDG activity optimized, the dose to the NMT is mainly caused by patient contact. Even if the patient has been instructed well, some coaching and body contact may be necessary, especially in the case of sick patients. Close patient contact is also responsible for the relatively high exposure of the radiotherapy technicians. Their dose is only slightly lower than that of the NMTs who perform a complete patient study.

The occupational exposure we realized is in the low range of what other investigators have reported [9 – 11], indicating that the working procedure is well optimized. Also in an absolute

sense an effective dose of 2.8 mSv per year, assuming a workload of 1000 patients each receiving 370 MBq $^{18}$F-FDG, seems quite acceptable. In fact, no statistical significant increase in readings of the legally obligatory personal dosimeters was observed after the introduction of PET/CT.

At this moment the department is working with a next generation PET/CT, the Philips Gemini TF, a system that uses time of flight detection. On this system the activity administered to the patient has been lowered from nominally 370 MBq to 180 or 220 MBq, depending on the body mass index of the patient. In addition, the activity required for the daily PET quality control is lower now: a 3.7 MBq $^{22}$Na point source instead of the bucket phantom with 15 – 40 MBq of $^{68}$Ge/$^{68}$Ga. Therefore, the exposure might now even be somewhat lower than during the initial optimization here described.

Finally one should consider exposure of persons outside the nuclear medicine department by a patient injected with $^{18}$F-FDG, who, as we have seen, is a non-negligible source of radiation. In a comprehensive study Cronin et al [15] investigated the exposure during travel, at the patient's work place, at home (partner and children in different age groups) and nursing staff: "The only possible area of concern is in an oncology ward, where patients may be regularly referred for PET investigations and other high activity radionuclide studies and are partially helpless. Even in this area, however, it is unlikely that a nurse would receive a daily dose of more than 24 μSv. We conclude that there is no need for restrictive advice for patients undergoing 18 FDG PET studies given the current administered activities." Cronin et al set a dose limit of 1 mSv/y for all persons, except for the nurses, for whom they applied 6 mSv/y. Using Table 6 from the article by Cronin et al. an estimate of the exposure of nurses in a local situation can be made.

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# Code of practice for individual monitoring of radiological workers with protective clothing

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NCS Subcommittee: Protocols for personal dosimetry of workers wearing protective clothing

Late 2005, the Dutch Ministry of Social Affairs and Employment (SZW) has requested the NCS to produce a code of practice to harmonise individual monitoring of workers in radiology when personal protective devices are employed. This paper explains the background of the code of practice and summarizes a few choices made from available alternatives.

Dutch legislation (Radiation Protection Decree, Bs2001 [1]) requires individual monitoring of workers who may be exposed to annual effective doses higher than 1 mSv. There are several reasons for individual monitoring, one of which is to assure that workers do not exceed the annual dose limit of 20 mSv. Usually, personal dose equivalent ($H_P(10)$, previously called depth dose) is obtained with a personal dosimeter (badge) based on thermoluminescence dosimetry (TLD). An approved dosimetric service (ADS), of which five exist in The Netherlands, issues dosimeters and takes care of reading them out periodically. The results in terms of $H_P(10)$ are registered in a centralised database. For this purpose the National Dose Registration and Information System (NDRIS) was established in 1989. Since then records of over 100,000 persons are kept [2]. In 2004 the number of registered workers in active service was almost 35,000. They belong to several categories distinguishing professional groups within branches, e.g. external therapy within health care, or isotope production within business. It is proposed to distinguish in future the type of employer, type of application and type of equipment used. University hospital, radiotherapy and linear accelerator is one example. Categorisation enables statistical overviews of professional exposure. NDRIS provides such reports periodically, e.g. for SZW. Employers also receive information from NDRIS about the exposure of their workers. Once entered into NDRIS a dose value cannot be changed easily. It requires good argumentation and approval by the Labour Inspectorate of SZW. Similar systems for individual monitoring exist in other European and North-American countries.

Dutch law prescribes monitoring of effective dose for radiological workers. Effective dose is a weighted sum of doses to radiosensitive organs. It is related to the stochastic risk due to exposure to ionising radiation, i.e. induction of fatal tumours and hereditary effects in offspring.

In practice it is impossible to measure effective dose directly. Instead, an operational quantity is measured. This operational quantity is obtained using a simple detector and is supposed to provide a conservative approximation of the desired parameter. The operational quantity is usually $H_P(10)$, defined by the International Commission on Radiological Units and Measurements (ICRU) [3] as absorbed dose to tissue at 10 mm depth below a specified point on the irradiated surface. Personal dosimeters are calibrated with respect to $H_P(10)$. Recent intercomparison studies concerning personal dosimetry [4] have shown that the variation in indication of equally exposed dosimeters mostly remains within ±20%. The irradiation fields in these cases were rather uniform and the photon energy not too low (> 40 keV). These conditions are good enough for radiation protection purposes as it is deemed acceptable when a measured dose value remains within a factor of 1.5 from the true dose.

In general, good correspondence of $H_P(10)$ and effective dose can be assumed for normal workplace conditions. However, the personal dosimeter must be worn at an appropriate location on the body, as there is considerable dependence on the irradiation geometry (anisotropy) and also on photon energy [5]. When the worker is wearing protective clothing, e.g. a lead-equivalent apron, an adequate indication of effective dose by a read-out personal dose value is no longer true. As the body is never shielded completely, a measurement below the apron yields too low a dose value, thus giving a wrongful sense of safety. In contrast, a dose measurement above the apron overestimates the dose as the protective effect of the apron is not taken into account. Based on latter types of reading, highly exposed workers may apparently exceed the annual dose limit. Such workers run the risk of suffering from consequences like unjust suspension of duties. Therefore, depending on the position of the dosimeter, some correction should be applied to the reading to achieve an appropriate assessment of $H_P(10)$ and effective dose.

In contrast to e.g. the UK, where a personal dosimeter is usually worn underneath the apron, in The Netherlands the wearing position is commonly outside the apron [6]. The latter situation has a number of advantages. Measured dose values will be in the range where instrumental accuracy is pretty good, whereas radiation intensity under the apron may stay below instrumental detection level. For the same reason it is easier to calculate correction factors for measurements outside the apron, e.g. using Monte Carlo simulation techniques. Furthermore, from a measurement outside the apron an indication of the dose to unshielded organs can be obtained, for instance the eyes.
Some countries (Belgium, Switzerland) prescribe double dosimetry for certain exposure conditions. It means that one dosimeter is worn underneath the apron, another one above it. An

algorithm that contains two correction factors can then be used to derive effective dose from both dosimeter readings. Multiple dosimeters obviously yield better information about the exposure, which can be used for radiation protection purposes. Double dosimetry, however, is error-prone because it is more complex (and it is also more expensive).
For standard procedures that at present are performed routinely in The Netherlands, a single personal dosimeter worn outside the apron is expected to be sufficiently adequate. For medical applications, a wearing position at a central location high on the chest seems most appropriate. The dosimeter can possibly be attached to a thyroid collar. Recently, Monte Carlo simulations for interventional cardiology were performed at Delft University of Technology. They have shown that this position seems to be least sensitive to variation of the direction of the primary beam.

Since 1993 a few ADS apply a correction, i.e. the dosimeter reading taken outside the apron is divided by a factor of 5, before providing the results to NDRIS. Other ADS report uncorrected results. Of course, differing policies are highly undesirable as it disturbs a fair comparison of the exposure of workers.

The NDRIS database reveals that the registered annual dose in 2004 exceeded 1 mSv, 6 mSv or 20 mSv for 1950, 185 or 23 workers (6%, 5‰, 1‰), respectively [6]. The majority of workers with a dose above 6 mSv belong to the medical professions (172, of which 164 in radiology and cardiology). This also holds true for the highest dose group. Only 2 persons out of 23 with doses above 20 mSv belong to other professional categories (mobile non-destructive testing).
Lead aprons are most often worn in hospitals and veterinarian practices. Frequency of use is 75% and 100%, respectively [6]. This has to do with the fact that the efficacy of lead aprons is significant in particular for the types of radiation and energy ranges used for medical imaging applications (50-140 kVp X-ray sources). Unwieldiness of the garment limits the thickness (attenuation effect) of the lead-equivalent material (0.15-0.7 mm Pb). In most other branches lead aprons are not worn at all, or the frequency is 20-30% at maximum.

Two groups of workers have been identified who wear lead aprons and, potentially, may be highly exposed because of their presence near the primary X-ray beam and in scattered radiation fields during prolonged periods of time. These are workers in interventional radiology and cardiology, and workers in veterinary diagnostic radiology. In the latter group high dose exposure is very exceptional; hence it seems appropriate to exclude veterinarians as relevant for the code of practice. Workers in interventional radiology and cardiology are the ones who

would really benefit from a more realistic estimation of effective dose by application of a correction factor to the dosimeter reading.

Applicable dose correction factors (CF) were collected from a literature survey and own calculations. Radiation quality (tube voltage 50-125 kVp), apron thickness (0.15-0.5 mm Pb), primary beam direction (antero-posterior, lateral), presence or absence of a thyroid collar, and dosimeter position (waist or neck, left-central-right) were varied. Values of CF ranging from 2 to 76 were found. This refers to the value by which the dosimeter reading, outside the apron, has to be divided to obtain the estimated effective dose.

The code of practice should balance accurate dosimetry and practical use. To obtain very realistic dose estimates a list of CF could be produced, which depends on several parameters. A rather practical simplified table was drafted instead (Table 1), with CF depending on apron thickness and thyroid collar only. This avoids a heavy administrative burden implied when multiple parameters, e.g. also tube voltage and beam direction, are taken into account. A disadvantage is that ignoring information about additional exposure conditions introduces more uncertainty, but considering all effects would be impossible anyway –e.g. usage of ceiling suspended lead-acrylic screens has not yet been evaluated.

Sacrificing (some) accuracy for simplicity, sufficient safety has to be put in. Application of the CF values in Table 1 will yield a conservative (overestimated) value of effective dose. The extent of overestimation, however, will be such that false indication of exceeding the annual dose limits becomes exceptional for staff performing current interventional procedures.

*Table 1: Correction factors (CF) without or with a thyroid collar as a function of apron thickness; divide reading of dosemeter worn outside the apron by CF to yield effective dose*

| apron thickness mm Pb equiv. | | NO thyroid collar | | WITH thyroid collar |
|---|---|---|---|---|
| 0.15* | | 5 | | 5 |
| 0.25 | | 5 | | 5 |
| 0.35 | | 5 | | 10 |
| 0.5 | | 10 | | 15 |

* Use of CF for 0.15 mm Pb equivalent apron thickness is **prohibited UNLESS** the tube voltage never exceeds 80 kVp (e.g. paediatric interventions).

Main points of the code of practice, for the Dutch situation, are summarized as follows.

- Concerns workers in interventional radiology and cardiology who routinely perform interventional procedures (X-rays with tube voltages up to 140 kVp) wearing a suitable lead apron.

- Worker wears the personal dosemeter on a central position high on the chest.
- Local expert in charge of the department's radiation safety has (Dutch) level 3 licence or better.
- Local expert and worker declare that work is performed in accordance with the code of practice.
- Local expert selects protective clothing and selects corresponding CF from Table 1.
- Local expert requires permission from Labour Inspectorate to have CF applied.
- ADS processes the personal dosimeter; applies CF; supplies corrected and uncorrected dosimeter reading + CF to NDRIS for registration.
- NDRIS administrator records the data and reports back in the usual manner.

At present (June 2007), the code of practice is still preliminary and has no official status as the corresponding NCS report could not yet be approved by the board of the NCS. Widespread acceptance and application of the code of practice by all parties concerned is highly desirable. Therefore, advice from several professional groups is still being awaited before publication. Official status also requires formal approval of SZW. Consequently, the code of practice cannot yet be applied.

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# Registration of radiation protection experts in the Netherlands

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## Introduction

In de European directive 96/29 Euratom [1] it is described that radiological activities have to be performed by or under the supervision of a qualified radiation protection expert. The qualified experts have to be registered. The European directive has to be implemented in the regulations of the member states of de European Community. This paper deals with the Dutch implementation.

## Implementation in the national regulations

*1. Implementation in the Netherlands*

In the Netherlands, European Directive 96/29/Euratom is implemented in the Radiation Protection Decree, Bs2001 [2], which came into act beginning 2003. With respect to the qualified expert, article 9 of the Bs2001 specifies that radiological activities must be performed by or under the supervision of a qualified radiation protection expert. Article 7 of the Bs2001 states that the qualified expert must be registered in a register appointed by the Ministers. Since then, a lot of work has been done and there have been many discussions as to how these general rules should be implemented into a workable system. This concerned the organisation of the system, the target group (who have to be registered) and the acceptance criteria for registration. These aspects are elaborated in sections 3 (organisation), 4 (target group) and 5 (criteria). To facilitate the process, information has been collected about the situation in some other member states.

*2. Implementation in some other member states*

In 2004, at the request of the Netherlands Ministry of Social Affairs and Employment, a brief review was carried out on the way the registration of qualified experts had been implemented by several member states of the European Community [3]. The result of this review showed major differences both in the registration system and the rate of implementation. Below, a short overview is given:

Germany

- All radiation protection experts working in the radiation protection organisation have to be qualified (certified) by a competent organisation (zuständige Stelle);
- Criteria for certification concern 'primary education', 'working practice' and 'radiation protection education';
- Certificate is valid for a period of 5 years;
- Total number of certified radiation protection experts is estimated to be substantially more than 20.000.

Belgium

- A company performing radiological activities doesn't need to employ a qualified radiation protection expert, but must have an agreement with a 'notified body'. Depending on the radiological risks, this must be a level 1 (high risks) or 2 (medium and low risks) notified body, which must be under leadership of a level 1 or 2 qualified expert, respectively;
- The criteria for recognition of experts mainly concern primary education and radiation protection education. For level 1 experts both must be at an university level, while for a level 2 expert a bachelor level suffices;
- Registration is performed by a governmental agency for periods of 5 years;
- In Belgium there are in total about 50 level 1 and 50 level 2 qualified experts.

Great Britain

- A company performing radiological activities doesn't need to employ a qualified radiation protection expert, but must have an agreement with a recognised Radiation Protection Officer (RPA);
- Criteria for recognition concern 'radiation protection education", 'working practice' and 'management and consultancy skills'. To fulfil these criteria, within each period of 5 years, a number of 'accreditation points' need to be collected, which can be obtained by following courses, attending symposia and congresses, writing and presenting papers, etc. Within each category a minimum number of points are needed;
- Recognition is performed by a recognised Assessing Body (the Society for Radiological Protection is one of them) for a period of 5 years;
- The total number of recognised RPAs in Great Britain is a few hundred.

Sweden

- At the time of the review (2004) there were only qualification criteria for Radiation Protection Advisers at nuclear power stations and for medical physicists. For other applications, recognition is carried out on an individual basis;
- The number of recognised experts (nuclear power stations and medical applications) at that time (2004) was a few tens.

**Organisation of the registration system**

The Netherlands has opted for a privatised system, in which the registration is performed by private organisations and the criteria are set by a committee of experts. The governmental role is to approve the criteria and to appoint the registration bodies. These registration bodies have to apply the criteria, which are set by the committee of experts. In figure 1 the organisation scheme is given.

An administration foundation manages both criteria for the experts themselves and for the registration bodies. These criteria are developed and kept state-of-the-art by a committee of experts, installed by the foundation. After approval of the criteria by the governmental authorities (in this case the Ministry of Social Affairs and Employment) one or more registration bodies are appointed. There are agreements between the different parties, to lay down obligations (e.g. financial and reporting). An administration foundation can act for more than one profession. In our case the foundation comprises the following related professions: radiation protection experts, safety experts, labour hygienists, and labour and organisation experts. For each profession there is a separate committee of experts. The foundation is in operation since the end of 2006.

It is expected that the organisation of the system will facilitate the possibility of combined registration and certifications for different professions (e.g. someone who has to be registered as radiation protection expert but also as safety expert or a clinical physicist who is also the responsible radiation protection expert).

**The target group (who has to be registered?)**

An exact interpretation of Bs2001 would result in the registration of some 15.000 experts [4], including dentists and veterinary surgeons. There is increasing doubt whether the registration of

such a large group is useful. This doubt is enhanced by the information obtained from other member states (see 2.2). Therefore, presently there are discussions as to what the target group should be. Ultimately, this is the responsibility of the Governmental authorities and a decision is expected soon. Below some insight in the ongoing discussion is presented.

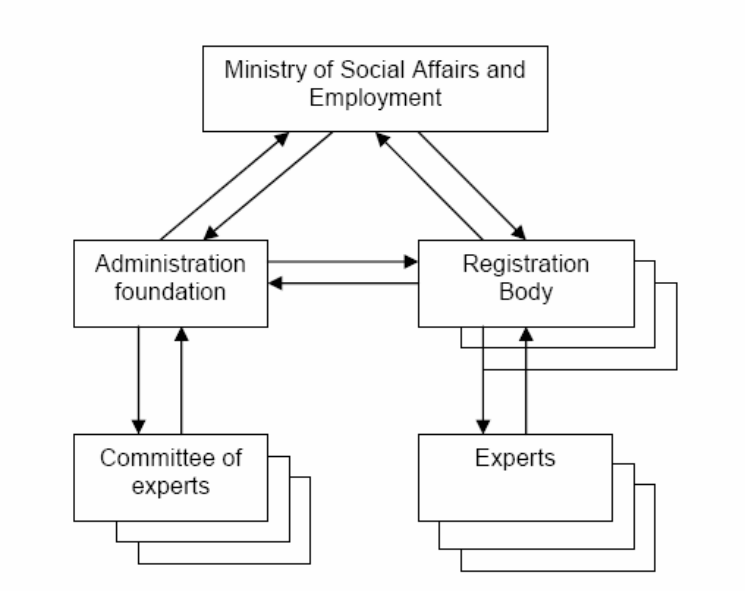


*Figure 1: Organisation scheme of the registration system*

In the Netherlands a comprehensive radiation protection education system has already been in operation for many years. In this system the following expertise levels are distinguished [4]:

Level 5A: Level of expertise needed for the use of enclosed radioactive materials and Roentgen apparatus involving small risks;

Level 5B: Level of expertise needed for the use of enclosed radioactive materials and open radioactive materials involving small risks;

Level 4A: Level of expertise needed for the use of all Roentgen apparatus as well as enclosed radioactive materials and other ionising radiation producing apparatus involving moderate risks;

Level 4B: Level of expertise needed for the use of enclosed radioactive materials and open radioactivity involving moderate risks;

Level 3: Level of expertise needed for the use of enclosed and open radioactive materials involving significant risks;

Level 2: Level of expertise, which is significantly higher than level 3 for the use of all enclosed and open radioactive materials as well as all ionising radiation producing apparatus;

Level 1: This level is in principle left open for experts operating at an internationally recognised level. However, there is and has never been any education for level 1, and in the Netherlands there are officially recognised level 1 experts.

The discussion with respect to the target group for registration is focussed on the level of expertise of the expert as well as on his/her responsibility in the radiation protection organisation.

Initially, it looked as if the discussions would lead to a decision by the authorities to restrict the requirement for registration to level 2, 3 and 4 experts responsible for licensed radiological activities. At present, however, it seems that also level 5 experts responsible for licensed radiological activities will have to be registered, but with less stringent criteria. This would reduce the target group from the original 15.000 down to about 1.000 or less (depending on the final decision of the governmental authorities). The final decision is expected at the end of 2007. It is likely that the results of the discussions within the European Platform on Training and Education in Radiation Protection [5] will be taken into account.

**Registration criteria**

A preliminary committee of experts has formulated a first set of registration criteria for the expertise levels 2, 3 and 4. The criteria deal with the following three aspects:

a. Initial radiation protection education;
b. Work experience;
c. Refresher courses and gaining knowledge.

Item a is fulfilled by passing the examination of a radiation protection course at the respective level. Required work experience is expressed in the mean number of working hours in relation to the application of ionising radiation over a period of five years. This criterion increases from 100 hr/y for level 4 to 250 hr/y for level 3 and 500 hr/y for level 2.

Within item c, ‘refresher courses’ and ‘ gaining knowledge are distinguished. A refresher course is intended to “refresh” the knowledge obtained during the initial basic education. Gaining knowledge relates to new knowledge that goes beyond the knowledge needed for passing the

examination of the initial course. Comparable to the system used in Great Britain, a certain number of accreditation points need to be collected during a registration period of 5 years. These points can be obtained by following refreshment courses and by gaining knowledge through attending symposia and congresses, writing papers, participating in committees, etc. The number of point needed increases with expertise level. For every category a minimum number of points is required. Some examples are:

- A level 4 expert needs 60 points in five years. This can be achieved by following a 1 day refresher course every year (5 x 10 points) and attending 2 symposia of one day (2 x 5 points);
- A level 3 experts needs 120 points in five years. This can be achieved by following a 2 days refresher course every year (5 x 20 points) and 4 symposia of one day (4 x 5 points);
- A level 2 expert needs 200 points in five years. This can be achieved by following a 2 days refresher course every year (5 x 20 points), attending 2 symposia of one day every year (5 x 10 points) and two congresses of 5 days (2 x 25 points).

These criteria might still be altered to some extent, in order to obtain some harmonisation between certification and registration criteria for the different professions participating in the foundation. The purpose for this is to facilitate combined registrations and certifications.

**Conclusions**

For some time, efforts have been made to set up a system for the registration of radiation protection experts in The Netherlands. Setting up this system is in its final stages, but still some final decisions have to be made with respect to the actual target group. Most probably the system will go in operation in 2008.

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