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Shortage of medical physicists, poor access to education and training and lack of professional recognition are issues prevailing in several countries. To address them, the IAEA carries out many wide-ranging activities in support of developments in medical radiation physics worldwide. They are mostly related to education and training, scientific guidance, research and development, and they also comprise dosimetry services.

In 2013, an IAEA report was published [1] that defined the roles and responsibilities of a clinically qualified medical physicist in radiotherapy, diagnostic radiology and nuclear medicine. The report, endorsed by the International Organisation for Medical Physics (IOMP) and the American Association of Medical Physicists (AAPM), also recommends requirements for the academic education and clinical training, and for continual professional development. The report helps to harmonize the approaches for education and clinical training across the world, as well as it promotes the recognition of medical physics as a profession. In 2015, participants from over 30 IAEA Member States in Europe gathered in Vienna to discuss the current status of and future perspectives for medical physics in the IAEA Europe Region. Representatives of international professional organizations, the World Health Organization, national regulatory bodies, Health Ministries and academia as well as medical physicists came together to discuss and build awareness of the important role of medical physicists in the practice of radiation medicine. Participants at the meeting endorsed a set of recommendations addressed to the IAEA Member States in the Region, which were disseminated by the IAEA and professional organizations. It is expected that the recommendations will bring progress in strengthening medical physics capacity in the Europe Region.

Education and training is a predominant IAEA activity area, which includes syllabi, publications aimed at education, training packages for teaching courses, and clinical training material. Three IAEA handbooks for teachers and students provide the basis for education of medical physicists initiating their university studies in radiotherapy, diagnostic radiology and nuclear medicine. These volumes are supplemented by a collection of lecture slides illustrating all handbook chapters. The IAEA teaching packages have been adopted by several universities and hospital training centres.

There is a growing awareness that radiation medicine is increasingly dependent on well-educated medical physicists appropriately trained in the clinical environment. However, the results of IAEA surveys suggest a significant shortfall of clinically qualified medical physics professionals, particularly in low and middle income countries. To address the gap, the IAEA developed three clinical training packages for the physics of radiation therapy, diagnostic radiology and nuclear medicine. Advanced Medical Physics Learning Environment (AMPLE) hosted by the IAEA’s CLP4NET also provides a structured clinical training guide that links to electronic resources for trainees, facilitates communication between trainees and
supervisors, and offers a grading system to allow the training progress to be recorded and tracked. To-date more than 100 medical physics trainees have been enrolled in AMPLE.

The IAEA has also published a number of technical reports, guidance documents and advisory books that are a useful resource for medical physics training, in particular for equipment acceptance testing and commissioning including imaging, treatment planning and record and verify systems, as well as for quality assurance (QA) in the various areas of clinical radiotherapy physics, both for equipment and patient related procedures. There are also numerous IAEA publications available in QA for imaging both in X-ray diagnostics and nuclear medicine. Multiple publications are available for medical physicists in specific areas of their activities on the IAEA ‘Human Health Campus’ [2] website which can serve as an educational resource for professionals and trainees.

For long time, the IAEA has maintained interest in standardization and development of codes of practice for dosimetry with several publications in the field. One important example is the TRS-398 [3] that is in use by medical physicists involved with radiotherapy dosimetry, and has been adopted by several countries as the national dosimetry protocol. A new code of practice for dosimetry of small radiotherapy fields, jointly prepared by the IAEA and the American Association of Medical Physicists is expected to be published soon.

There is also a long tradition of organizing conferences and symposia in dosimetry and medical radiation physics to foster the exchange of information among professionals and to highlight new developments. The last International Symposium on Standards, Applications and Quality Assurance in Medical Radiation Dosimetry (IDOS) was organized by the IAEA in 2010. The symposium provided a forum for discussion of advances in radiation dosimetry made during the previous decade with a specific focus on the challenges of dosimetry in small and non-standard radiotherapy beams. The symposium also outlined trends in medical radiation dosimetry, and identified possible areas for developments. IDOS-2 will take place in 2019 and it will overview developments of the last decade.

Regional and national training courses, workshops scientific visits and fellowships in recognized radiotherapy centres abroad are available for medical physicists to upgrade their knowledge and skills through the IAEA Technical Cooperation (TC) projects. Within the European TC projects, hundreds of medical physicists received support for the participation in teaching courses organized by the European Society for Radiotherapy and Oncology (ESTRO) and InHolland University. Using the IAEA TC support, the Association of Medical Physicists in Russia has established a training hub in Moscow to address the educational needs of Russian speaking medical physicists. Another opportunity to participate in IAEA supported training is through teaching courses jointly operated by the IAEA and the International Centre of Theoretical Physics (ICTP) in Trieste, Italy. These courses cover various topics in QA, imaging and radiotherapy physics. Recently, the Master of Advanced Studies in Medical Physics (MMP) [4] was developed at ICTP that provides physics graduates with education and clinical training so that they may be recognised as clinical medical physicists in their home countries. The MMP programme has been accredited by IOMP and is co-sponsored by the IAEA. Also, there exists the ICTP-IAEA Sandwich Training Educational Programme (STEP) that offers fellowship opportunities to PhD students in low and middle income countries.
Research opportunities exist through the IAEA Coordinated Research Programme (CRP) involving topical multicentre studies. Examples include developing methodologies for advanced dosimetry audits in radiotherapy, studying accuracy and uncertainties in medical physics aspects of radiotherapy, and testing a new code of practice for small beam dosimetry. Also, doctoral CRPs are available that couple PhD students from low and middle income countries with mentors from advanced radiation medicine centres. The current doctoral CRP embraces topics related to quality assurance in diagnostic radiology.

The IAEA support to medical physics also includes dosimetry services. Traceable dosimetry calibration services are provided by the IAEA Dosimetry Laboratory through the IAEA/WHO Network of Secondary Standard Dosimetry Laboratories (SSDLs) to promote accurate measurements of radiation doses. Inter-laboratory comparisons of ion chamber calibrations and dose quality audits using mailed solid state dosimeters are also offered. Both programmes are available to SSDLs, while dose quality audits are provided to radiotherapy centres through the IAEA/WHO postal dose programme.

Over 100 calibration coefficients are determined yearly by the IAEA Dosimetry Laboratory for national secondary standards for radiotherapy dosimetry. About 40 SSDLs participated in in the inter-laboratory comparison programme with all recent results within the acceptance level of 1.5%. The TLD audit programme for SSDLs annually checks about 30-40 beam calibrations with >98% results within the acceptance level of 3.5%. Any discrepancies are followed-up and corrected. These programmes confirm that the majority of SSDLs are capable of disseminating dosimetry standards to hospitals with the acceptable uncertainty. The IAEA/WHO postal dose audit service for radiotherapy hospitals has checked approximately 12300 radiotherapy beam calibrations in 2230 hospitals in 132 countries. At present, about 97% results are within the acceptance level of 5%, compared to 78% in the year 2000. Subsequent follow-up actions of poor results have helped many radiotherapy centres to resolve the discrepancies in dosimetry, thus preventing potential dose misadministration to cancer patients. Generally, the basic dosimetry practices improved significantly over the years; however, in a few countries they are not fully satisfactory yet.

Overall, multiple IAEA projects are in operation at various levels, which help in the development and growth of the medical physics profession worldwide.

References


2. IAEA Human Health Campus: https://nucleus.iaea.org/HHW/index.html


Introducing a new radiophotoluminescent dosimetry (RPLD) system for the IAEA/WHO postal dose audits in radiotherapy

Paulina Wesolowska, Tomislav Bokulic, Pavel Kazantsev, Joanna Izewska

*International Atomic Energy Agency, Vienna, Austria*

j.izewska@iaea.org

Introduction

The IAEA Dosimetry Laboratory (DOL) helps to improve the accuracy of clinical dosimetry in radiotherapy centres world-wide by providing independent dose audits of radiation beams used for cancer treatment. A thermoluminescent dosimetry (TLD) system has been used by DOL for its auditing programme for over 47 years [1]. A new radiophotoluminescent dosimetry (RPLD) system was introduced in 2017 to replace the aging TLD equipment. Prior to its use for audits, the RPLD system underwent a thorough commissioning process. The RPLD system’s parameters, readout methodology and the dosimetric characteristics were investigated in depth and compared to the well-established TLD system characteristics.

Material and Methods

The IAEA RPLD system consists of GD-302M glass rods and a FDG-1000 Dose Ace reader by Asahi Techno Glass Corporation (ATG). The glass rods are made of silver activated phosphate glass; they are 12 mm long and 1.5 mm in diameter. The sensitive area of a dosimeter is 6 mm long. RPLDs are encapsulated in custom made watertight capsules. The RPLD readout is fast; 20 glass rods can be loaded at a time to the reader for the readout session of about 5 min. RPLDs can be re-read several times as the readout process is non-destructive. Dosimeters can be reused after annealing.

To determine RPLD characteristics, the dosimeters were irradiated with two Cobalt-60 units at DOL (Nordion X-200 and Picker V40) and also with an Elekta Synergy accelerator at the Medical University of Vienna. Prior to each dosimeter irradiation, a dose determination was performed with a Farmer type ionization chamber with the calibration coefficient traceable to the Bureau International des Poids et Mesures (BIPM).

The absorbed dose can be determined from the dosimeter reading using the following equation:

\[ D = M \times SCF \times N \times f_{lin} \times f_{en} \times f_{hol} \times f_{fad} \]

where \( M \) is the dosimeter reading (corrected for the depletion and the readout position, see below), SCF is the dosimeter sensitivity correction factor, \( N \) is the dosimetry system calibration coefficient, \( f_{lin} \) is the dose response non-linearity correction factor, \( f_{en} \) is the energy correction factor, \( f_{hol} \) is the holder correction factor and \( f_{fad} \) is the fading correction factor.

Individual dosimeter sensitivity correction factors (SCFs) have been determined by grouping RPLDs in batches of 100 dosimeters. Each group was irradiated in a Co-60 beam with 2 Gy dose. The readout of a batch of 100 dosimeters was performed in one session without breaks between readouts. SCFs were calculated as the ratio of an average signal of the 100-dosimeter batch to the signal of an individual dosimeter within the batch. For readout, RPLDs are placed in a tray that could accommodate up to 20 dosimeters. As the dosimeter response varies with its position in the readout tray, the readout position corrections were determined. Partial loss of the signal (depletion) occurs for every readout and the repeat readouts were...
used to determine the depletion effect. As the time gaps between the readouts affect depletion, additional tests were performed to examine the effect in more detail.

To study the readout reproducibility, four repeat readouts of the same dosimeter were performed for 2481 dosimeters. The relation between the dose delivered and the RPLD signal measured was determined in the dose range of 1 to 4 Gy and corrections for the dose response non-linearity were established. The energy response was determined by comparing RPLD signals for dosimeters irradiated to 2 Gy dose with high energy photon beams of 6 MV–18 MV and with a Co-60 beam. The holder correction was determined from the integrals of the area under the beam profiles within the sensitive volume of RPLD with and without the holder in place. The profiles were measured in water, using a microdiamond dosimeter. Fading was studied by reading the signal of 20 RPLDs irradiated with 2 Gy in a Co-60 beam, in regular intervals over the period of three months.

The overall combined standard uncertainty was estimated based on uncertainties of individual components in the equation above.

Finally, a multicentre pilot audit study was performed using TLDs and RPLDs irradiated with 2 Gy dose in the same measurement sessions with 20 high energy photon beams including Co-60 and 6 MV–18 MV beams by seven radiotherapy centres in Australia, Austria, Hungary, Singapore, Slovakia, Sweden and USA.

**Results**

The readout reproducibility calculated for 2481 dosimeters was 0.16%, determined as the standard deviation of the mean for the series of four repeat readouts of the same dosimeter. Depletion of 0.017% per readout was established from a session of 200 readouts with no time delay between the readouts. The standard RPLD readout procedure includes a time gap of 5 minutes between the readouts. The depletion measured for the standard readout procedure was 0.008% per readout. The depletion for different readout sequences is shown in Fig. 1. It was observed that the signal returns to its initial value after a time gap of the order of days. The corrections for the RPLD readout position was within ±0.5% for all dosimeter positions in four readout trays used at DOL.

The distribution of SCFs determined for 4980 RPLDs has the standard deviation σ=0.8% with all values falling between 0.954 and 1.062.

In Fig. 2 the RPLD dose response is compared to TLD dose response in the range of doses relevant to their intended use, i.e. 1.5–2.5 Gy.

The results of energy dependence study resulted in the energy corrections for high energy photon beams of 6 MV–18 MV of 1.020–1.032 and 1.015–1.024 for TLDs and RPLDs, respectively, with the normalization point of $f_{en}=1$ for a Co-60 beam.

For RPLDs, the correction for attenuation and scatter in the IAEA standard holder ranged from 1.007 to 1.006 for Co 60 and 18 MV beams, respectively. For TLDs, these values were 1.018 and 1.007, respectively.

The RPLD fading effect showed about 0.4% signal loss after one hundred days. However, if RPLDs are read within two weeks post irradiation, the fading effect becomes negligible.

The summary of RPLD uncertainties are given in Table 1 together with TLD uncertainties. The quadratic summation of all uncertainty components leads to the combined standard uncertainty of 1.5% for RPLDs and 1.6% for TLDs.
To conclude the RPLD system commissioning, a multicentre pilot audit study was performed using TLDs and RPLDs irradiated in parallel. The results are given in Figure 3. The distribution of the TLD results gives the mean of 1.001 and the standard deviation of 1.6%, whereas RPLDs results have the mean of 0.998 and the standard deviation of 1.1%. The differences between the TLD and RPLD determined doses for individual data points were within 1.5%.

Discussion

The RPLD system commissioning involved several steps including the determination of the system calibration and a set of correction factors to account for different effects, such as the individual dosimeter sensitivity, the dose response non-linearity, the energy correction, fading and the holder effect. The results were compared with literature as well as the parameters of the well-established TLD system.

The distribution of SCFs demonstrates that 4980 dosimeters tested in this study have consistent sensitivities with \( \sigma = 0.8\% \). Nevertheless, for accurate dosimetry SCFs need to be applied in the dose calculation. It can be seen from Figure 2 that the RPLD dose response is sub-linear whereas TLD response is supra-linear in the dose range of interest. Larger dose-response nonlinearity corrections are necessary for TLDs than for the RPLDs. Rah et al. [2] showed similar dose response characteristics for TLDs and RPLDs. Mizuno et al. [3] reported comparable changes in dosimeter responses to high energy photon beams for RPLDs and TLDs. The holder correction for RPLDs was found lower than that for TLDs due to different dosimeter volumes affected by attenuation and scatter of radiation by the holder. Rah et al., [4] reported greater signal fading of 1.6\% after 133 days compared to 0.4\% after 100 days in this work. The uncertainty of 1.1\% for the RPLD system used for dosimetry audits in Japan was reported by Mizuno et al. [3] which is tighter than reported in this work (1.5\%).

The results of the multicentre pilot audits for RPLDs and TLDs irradiated in parallel were consistent with each other. Overall, it was found that RPLD system has appropriate characteristics for auditing purposes, with the uncertainty levels comparable or lower than those for TLDs.

Acknowledgements: The IAEA consultants: H. Mizuno, T. Fujibuchi, I. Diallo and interns: B. Bencsik, T. Santos, D. Szegedi contributed to the RPLD system commissioning. Their contributions are gratefully acknowledged.

References

FIG. 1. RPLD signal depletion for various reading sequences.

FIG. 2. Dose response non-linearity for TLDs (red squares) and RPLDs (green triangles).
FIG. 3. Results of the multicentre pilot study with TLDs (blue diamonds) and RPLDs (red circles) irradiated in parallel with the same beams.

TABLE 1. Uncertainties in the dose determination from TLD and RPLD readings.

<table>
<thead>
<tr>
<th>Uncertainty component</th>
<th>TLD system</th>
<th>RPLD system</th>
</tr>
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<tbody>
<tr>
<td></td>
<td>Uncertainty (k=1) / %</td>
<td>Uncertainty (k=1) / %</td>
</tr>
<tr>
<td></td>
<td>Typ e A</td>
<td>Typ e B</td>
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<tr>
<td>Calibration of the dosimetry system</td>
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<tr>
<td>Determination of Co-60 dose from ionisation chamber readings</td>
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<td>Phantom positioning during irradiation</td>
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<tr>
<td>Solid water to water dose correction</td>
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<td>0.50</td>
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<tr>
<td>Dosimeter positioning during irradiation</td>
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<td>0.12</td>
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<td>Dosimeter readout</td>
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<td>0.15</td>
</tr>
<tr>
<td>Individual dosimeter sensitivity factor</td>
<td>-</td>
<td>0.42</td>
</tr>
<tr>
<td>Dosimeter positioning during readout</td>
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<td>0.42</td>
</tr>
<tr>
<td>Source of Uncertainty</td>
<td>Uncertainty (k=1)</td>
<td></td>
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<tr>
<td>-----------------------</td>
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<td></td>
</tr>
<tr>
<td>Combined standard uncertainty (k=1)</td>
<td>0.60 0.50 0.80 0.54</td>
<td></td>
</tr>
</tbody>
</table>

**Determination of the absorbed dose from dosimeter readings**

<table>
<thead>
<tr>
<th>Source of Uncertainty</th>
<th>Uncertainty (k=1)</th>
</tr>
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<tr>
<td>Calibration of the dosimetry system</td>
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<tr>
<td>Individual dosimeter sensitivity factor</td>
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</tr>
<tr>
<td>Dosimeter positioning during readout</td>
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</tr>
<tr>
<td>Non-linearity dose response correction factor</td>
<td>0.90 0.55</td>
</tr>
<tr>
<td>Energy correction factor</td>
<td>0.95 0.81</td>
</tr>
<tr>
<td>Fading correction factor</td>
<td>0.02 0.01</td>
</tr>
<tr>
<td>Holder correction</td>
<td>0.10 0.14</td>
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<tr>
<td>Combined standard uncertainty (k=1)</td>
<td>1.60 1.51</td>
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</table>
The Italian inter-society guidelines for radiation protection optimization in interventional radiology

R. Padovani,

_Abdus Salam International Centre For Theoretical Physics (ICTP), Trieste, Italy,_

To present the Italian Guidelines for the optimization of patients and staff radiation protection in interventional radiology, recently developed by a Working Group promoted by the Italian National Institute of Health and the National Workers Compensation Authority and consisting of 11 Italian scientific/professional societies involved in the fluoroscopically guided interventional practices.

Radiologists, interventional radiologists, neuroradiologists, interventional cardiologists, occupational health specialists, medical physicists, qualified experts, radiographers and nurses were represented in the Working Group.

The subject was divided in 3 main sections: radiation protection of patients (summarized in 10 “golden rules”); radiation protection of operators (other 10 “golden rules”); education/training of medical professionals.

In the “golden rules” practical and operational recommendations were provided that could help the professionals to reduce the dose to themselves by reducing the dose to patients too. Guidelines provided also indications about continuing education and training, and recommendations on accreditation and certification of professionals.

The methodology adopted for the development of these Guidelines with the contribution and agreement of all involved professionals can be seen as the winning approach for the distribution and practical implementation of the recommendations to reach a real impact on the optimization of the interventional radiology practices.
Training Opportunity for Young Physicists from Developing Countries at ICTP

R. Padovani¹, R. Longo², L. Bertocchi¹, M. De Denaro³,
¹ Abdus Salam International Centre For Theoretical Physics (ICTP), Trieste, Italy, ² Dept. Of Physics & INFN, Trieste University, Trieste, Italy, ³ Medical Physics Dpt, University Hospital, Trieste, Italy,

The Abdus Salam International Centre for Theoretical Physics (ICTP) and the Trieste University have initiated in 2014 a Master of Advanced Studies in Medical Physics (www.ictp.it/programmes/mmp.aspx), a two-years training programme in Medical Physics, co-sponsored by the Trieste University Hospital. The Master Programme is designated to provide young promising graduates in physics, mainly from developing countries, with a post-graduated theoretical and clinical training suitable to be recognised as Clinical Medical Physicist in their countries.

Presently, the 4 cycles of the Master programme has seen 80 participants from 44 Countries mainly from Africa, and Central and South America, selected among more than 1000 applicants. Full or partial scholarships are awarded to successful candidates from developing countries, thanks to the support of the IAEA, KFAS, IOMP, EFOMP and ICTP.

The Master programme has been developed according to the recommendations of IOMP and IAEA for the education and the clinical training. In the first year 332 hours of lectures and 228 hours of guided exercises are devoted to all main fields of medical physics. The second year is spent in one of the 19 medical physics department of the hospitals’ network for a full time clinical training in two areas: radiation oncology and diagnostic and interventional radiology and nuclear medicine.

Other training activities are the College of Medical Physics, a biannual event addressed to young physicists aiming to be training in the field of diagnostic and nuclear medicine imaging. Last September the 13th edition has taken place with the about 50 participants.

More recently a two-week training course on radiation therapy has been initiated with the support of IOMP, AAPM, EFOMP and AIFM. Last March-April the 3rd school has seen the participation of about 40 medical physicists.

Advanced training course are also jointly organised with IAEA. These are usually a week workshop with a narrow topic that see the discussion on the most advanced topics of medical physics, like, between others, internal dosimetry for therapeutic applications of radiopharmaceuticals, monte carlo techniques for medical applications, optimisation of CT procedures.

Medical physics community, IOMP and IAEA are seeing these initiative as an answer to the growing demand of training and continuous education Medical Physicists in developing Countries. Tanks to the synergy of ICTP, an international institution, Trieste University medical physics research group and several medical physics hospital departments in Italy and Croatia, these initiatives represents an important international contribution to the development of medical physics in the developing world.
Preparing young Medical Physicists for future leadership roles in Europe: an update

Carmel J. Caruana,

Prof. and Head, Medical Physics Department, Fac. of Health Sciences, University of Malta
EFOMP Past-Chair, Education and Training
EFOMP rep European Guidelines on the MPE, MEDRAPET and EUTEMPE-RX projects
carmel.j.caruana@um.edu.mt

Virginia Tsapaki

EFOMP projects chair, IOMP Secretary General

Introduction

Preparing future leaders has become an important issue in all professions – it is particularly crucial for small professions such as the Medical Physics profession. In particular Medical Physicists at expert level (Medical Physics Experts) need to develop leadership qualities if they are to be successful in motivating staff in their departments and ensure that their departments are successful. In his time as EFOMP Chair for E&T, the author realized that if the profession is to move forward we need to have a formal E&T process for preparing the next generation of leaders.

Materials and Methods

A literature review was carried with respect to leadership in the health care professions. Course modules in leadership particularly for the medical professions and specialties were analyzed and elements of best practice noted. These were then applied to the case of medical physicists.

Results

The EUTEMPE-RX module ‘Module MPE01: Leadership in Medical Physics: Development of the profession and the challenges for the MPE (D&IR)’ was developed and can be considered as a mini-MBA for Medical Physicists. Table 1 shows the learning outcomes whilst Table 2 some sample exam questions. It has been delivered twice (2014-15, 2016-17), combines online and onsite learning (Prague) and has been highly successful. It is now also attracting participants from outside Europe (presently 2 from US, 1 from Chile, 2 from Sudan).
Discussion

A similar module is now being developed by the EUTEMPE-RO consortium for Medical Physicists in Radiation Oncology under Horizon 2020 and one in nuclear medicine EUTEMPE-NM is also planned.

<table>
<thead>
<tr>
<th>Learning Objectives for EUTEPME-RX Module MPE01</th>
</tr>
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<tbody>
<tr>
<td>MPE01.01 Take responsibility for researching, evaluating, leading, and offering vision for the development of the role of the MPE (D&amp;IR,) in the ambit of European and national legislation and a holistic vision of healthcare.</td>
</tr>
<tr>
<td>MPE01.02 Implement and evaluate strategic solutions to the challenges facing the MPE (D&amp;IR) in own country and Europe.</td>
</tr>
<tr>
<td>MPE01.03 Evaluate the various models of management in terms of suitability for a Medical Physics Service and the issue of staffing levels.</td>
</tr>
<tr>
<td>MPE01.04 Take responsibility for the development of the role of the MPE (D&amp;IR) in health care governance and management in D&amp;IR.</td>
</tr>
<tr>
<td>MPE01.05 Take responsibility for ethical issues in medical physics particularly in the areas of research and radiation protection in D&amp;IR and apply them in practice.</td>
</tr>
<tr>
<td>MPE01.06 Discuss the role of the MPE (D&amp;IR) in service development, health technology assessment (HTA), innovation and expert consultancy.</td>
</tr>
<tr>
<td>MPE01.07 Research, develop and lead the development of the role of the MPE (D&amp;IR) in the education and training of medical physics trainees and other healthcare professionals.</td>
</tr>
<tr>
<td>MPE01.08 Manage the relationship of the MP/MPE with other healthcare professions in D&amp;IR, with patients and with the general public.</td>
</tr>
<tr>
<td>MPE01.09 Manage priorities regarding radiation protection research and medical physics input to clinical research projects needing the support of MPEs.</td>
</tr>
<tr>
<td>MPE01.10 Implement safety culture in their practice, participate in a clinical audit.</td>
</tr>
<tr>
<td>MPE01.11 Participate in networks for research and development at the European and international level.</td>
</tr>
<tr>
<td>MPE01.12 Take responsibility for the role of the MPE (D&amp;IR) in accidental and unintended medical exposures in D&amp;IR.</td>
</tr>
<tr>
<td>MPE01.13 Interpret the significance of liaising with the Radiation Protection Expert</td>
</tr>
</tbody>
</table>
### Table 2 Sample examination questions

| Case Study 1: Up to now there have only been Medical Physics Experts in Radiation Oncology and Nuclear Medicine in your country. However, EU Directive 2013/59/EURATOM has recognized the importance of the MPE also in Diagnostic and Interventional Radiology. You are having discussions about this issue with your healthcare authorities. One representative from the Ministry of Health tells you: “I can’t understand why Medical Physicists are required in Diagnostic and Interventional Radiology. In addition, you don’t have the high doses you have in Radiation Oncology” How would you tackle it? |
| Case study 2: There are 5 chest radiography rooms in your hospital each run by a different team of radiographers. You have noticed that one of the rooms is repeatedly exceeding the local DRLs which you have established. How would you tackle it? You know that the team of radiographers doesn’t like people investigating their techniques. |
| Case study 3: You are the head of the Medical Physics department at a large hospital which is expanding its Diagnostic and Interventional facilities owing to a large population increase in the region. You want to employ additional medical physics staff but the human resources manager tells you that you have enough. How would you tackle it? |

### References


The present and future role of Diagnostic and Interventional Radiology and the role of Medical Physicists

Carmel J. Caruana,

*Prof. and Head, Medical Physics Department, Fac. of Health Sciences, University of Malta EFOMP Past-Chair, Education and Training EFOMP rep European Guidelines on the MPE, MEDRAPET and EUTEMPE-RX projects*

carmel.j.caruana@um.edu.mt

**Introduction**

To be able to develop our role as Medical Physicists in Diagnostic and Interventional Radiology (D&IR) we must not only look at our role as defined in 2013/59/EURATOM and as elaborated in the ‘European Guidelines on the MPE’ *but also on the present and future role of D&IR itself*. This paper presents the results of a literature review and analysis of legislation/documentation regarding the present and future role of D&IR in healthcare and its possible impact on the model for subspecialization of the Medical Physicist in D&IR.

**Materials and Methods**

An analysis was carried out of articles and documents on the present and future role of D&IR in health care in the main radiology journals and professional websites using the search phrases ‘role of radiology’ and ‘future of radiology.’

**Results**

*Present role of D&IR in health care*

D&IR today involves an ever-expanding set of devices and techniques for investigating pathology, health screening and modifying non-invasively or with minimal invasion the morphology and functioning of the various parts of the human body. D&IR devices are also used extensively outside the D&IR department proper: cardiology, orthopaedic surgery, obstetrics, pulmonology and oral and maxillofacial surgery among others. Interestingly, 2013/59/EURATOM has also introduced the area of *non-medical imaging* exposure where non-medical imaging exposure means “any deliberate exposure of humans for imaging purposes where the primary intention of the exposure is not to bring a health benefit to the individual being exposed“ (2013/59/EURATOM Art 4). An indicative list of such practices is given in Annex V of the directive.

*The future development of D&IR in health care*
The present issues influencing the future development of D&IR and hence the role of the various professional groups working within it are:

(a) Information overload required of D&IR professionals in secondary, tertiary and quaternary healthcare institutions owing to:

- The rapid increase in number and sophistication of D&IR devices;
- The move away from strict morphological imaging to expansion into biomolecular and functional imaging for the earlier detection of disease at the cellular level;
- The rapid increase in number and sophistication of procedures in anatomical, functional, molecular, spectroscopic imaging;
- The increasing importance of quantitative imaging;
- The ever increasing accumulation of research data which fails to transfer to the clinic;
- The fact that medical specialists in secondary, tertiary and quaternary healthcare institutions are specialists in particular body systems / regions and D&IR professionals need to be very knowledgeable in such medical specialties in order to be able to liaise effectively with these medical specialists;
- The continuously increasing information regarding justification and dose optimization techniques;
- The increasing amount of legislation, guidelines and recommendations with which such professionals are expected to comply.

(b) The increased use of D&IR devices outside the D&IR department proper and the need to collaborate with and to provide expert advice to other departments in particular:

- The use of projection imaging guided procedures in cardiology, orthopaedic surgery, obstetrics and oral and maxillofacial surgery among others.
- The use of CT and MRI in mixed-modalities in nuclear medicine departments (PET-CT, PET-MRI, SPECT-CT etc devices);
- The ever increasing role of imaging devices in cancer detection and staging and treatment planning and monitoring;
- Use of D&IR devices in health screening institutions (e.g., breast screening units);
- Provision of expert consultation to primary care healthcare professionals on appropriate D&IR procedures for their patients. Before the rapid expansion of D&IR most primary care physicians in direct contact with the patient understood which specific D&IR studies were required for each condition. However, today, the primary care physician is overwhelmed by the ever increasing multitude of diagnostic imaging tests becoming available. Today, most general practitioners, internists, and paediatricians routinely consult with D&IR departments.
Other issues include the increasing awareness of the importance of radiation protection of the patient and special groups e.g., paediatric patients and the rise of radiogenomics.

**Discussion**

In this discussion we discuss the impact on the above developments on the subspecialist role of the various professions within D&IR.

**Radiologists:** When D&IR devices were restricted to projection x-ray devices and simple CT scanners and the number of procedures few it was possible for the General Radiologist to master the application of these devices to all body systems/regions and at all levels. It is being increasingly recognised that the General Radiologist who today tries to be an expert for all modalities and all body systems/regions severely risks being an expert in none. There is therefore an ongoing gradual demise of the traditional General Radiologist in secondary, tertiary and quaternary healthcare institutions. The question has therefore arisen on whether subspecialisation of Radiologists should be by modality or body system/region? Radiologists are moving to subspecialisation based on body system/region since to create added value for the referring healthcare professional the Radiologist needs to be fully knowledgeable of the body system/region and understand the clinical problem. The “General Radiologist” is being replaced with “collaborating subspecialty Radiologists”. The ESR (2011) has declared officially that the “The Society supports organ- rather than technology-based subspecialisation, but is cognisant that with increasingly sophisticated and complicated techniques, a “disease”-oriented educational matrix will inevitably be required in certain areas” and this is reflected in the ESR’s European Training Charter for Clinical Radiology.

**Diagnostic Radiographers:** Most Diagnostic Radiographers subspecialize by imaging modality and most MSc in Radiography are modality based. Some are body system/region based e.g., mammography, vascular ultrasound but again mostly focussed on a single modality. Single modality MSc are more attractive as they are easier to achieve.

**Medical Physicists:** The question arises whether in D&IR we should specialize by imaging modality (as in the case of Radiographers) or body system/region (as in the case of Radiologists) or perhaps a hybrid of both. An interesting hybrid model is being developed at Karolinska Institutet Stockholm and will be discussed.

**References**


Parizel P. (2017 ESR President) provides indication of radiology's future.
5
Verification of advanced radiotherapy techniques

Eduard Gershkevitsh

North Estonia Medical Centre

Advanced radiotherapy techniques (IGRT/IMRT/VMAT/SRS/SBRT/gating/tracking) are becoming a standard treatment for cancer patients in many countries. While significant technological progress is made to streamline the implementation and delivery of advanced techniques, the quality and safety of treatment depend on the availability of comprehensive QA procedures. Different methods (2D arrays, films, EPID, virtual software, etc.) exist to perform the verification of dose delivery and equipment performance. The review of these methods will be presented, and the strengths and weaknesses for each will be discussed.

Learning objectives:
To understand the challenges of advanced radiotherapy
To learn which parameters need to be checked
To familiarise themselves with methods and equipment used for verification
Alpha particle emitter therapy: review and practical examples

Mario de Denaro

Medical Physics Department - Azienda Sanitaria Universitaria Integrata di Trieste (ASUITs)
mario.dedenaro@asuits.sanita.fvg.it

Introduction
The employment of radionuclides emitting alpha particle for therapy has been investigated for many decades since the early1900s. The advantage in using alpha emitters in cancer treatment arises from the densely ionizing track and short path length [1]. For most of the investigated therapeutic alpha radionuclides the linear energy transfer (LET) ranges around 100 keV/μm, whereas for therapeutic beta radionuclides LET ranges around 0.2 keV/μm. High-LET radiation causes much more breaks in double-stranded DNA than low-LET radiation, because the maximum DNA breaks occur with LETs of 100-200 keV/μm [2]. The isotopes 224 and 226 of Radium were widely employed in the early use of alpha emitter radionuclides for several medical applications, showing dramatic late side-effects. The first clinical trial to investigate targeted therapy using alpha emitters in humans was carried out in 1997 [2] employing $^{213}$Bi. The use of monoclonal antibodies as carriers opened other promising perspectives in the field of radioimmunotherapy by beta particle emitters as well as by alpha emitter radioisotopes like $^{211}$At and $^{225}$Ac. Recently the Food and Drug Administration (FDA) approved the clinical use of the alpha-particle drug Radium 223 dichloride [3], for the treatment of patients with bone metastases. Another recent field of research on alpha particle emitter therapy, known as Boron Neutron Capture Therapy (BNCT), investigates the possibility to produce alpha radiation by nuclear reaction directly inside the patient, irradiating with thermal neutrons non-radioactive Boron 10 injected into the patient in the form of delivery agents for selective tumor targeting.

Material and Methods
Two examples of therapy connected to alpha particle emitter will be reported. The first one consist in the treatment of patients with $^{223}$Ra dichloride drug: the description of the management of this radioisotope in a Nuclear Medicine Department is detailed. The second one, oriented to the future of BNCT, consist in a research study for the implementation of a device named PhoNeS [4] capable of generate thermal neutrons by means of mega voltage Varian Clinac 2100 linac available in our hospital.

Results
Due to the half life of 11.4 days and because of its bone-seeking properties, $^{223}$Ra is a radioisotope particularly suitable for bone metastases therapy. The presence of photon emission in the range of hundreds of keV allows the detection with the same instrumentation used in a nuclear medicine department, facilitating the radiation protection procedures. The introduction of $^{223}$Ra therapy in our hospital required several steps: a formal authorization from regional authority, setup of the activity calibrator, setup of the monitoring systems for workers safety, waste disposal and training of the operators. Regarding our experience...
related to the BNCT, in our hospital we carried out some test fixing the PhoNeS device in the head of a Varian Clinac 2100. For different MU PhoNeS was able to produce neutrons flux with epithermal and thermal energies that was measured by bubble detectors.

Discussion
Alpha particle emitters therapy represents a very important field in the future of cancer therapy, even if further advances will be required in pharmacokinetics and in the methods of isotope production. We report our experience related to the introduction to our hospital of the therapy with $^{223}$Ra dichloride, an officially FDA approved drug. Another study without patients, at research level, was carried out in our institute to evaluated the capability to produce thermal neutrons by a conventional clinical linac, in order to evaluate possible future implementations of boron neutron capture therapy directly in the hospitals.

References
European epidemiological study on radiation induced lens opacities among interventional cardiologists

Olivera Ciraj-Bjelac\textsuperscript{1}, Lara Struelens\textsuperscript{2}, Anssi Auvinen\textsuperscript{3}, Peter Covens\textsuperscript{4}, Ulrike Scheidemann-Wesp\textsuperscript{5}, Alfred Wegener\textsuperscript{6}, Anders Widmark\textsuperscript{7}, Joanna Domienik\textsuperscript{8}, Octavian Dragusin\textsuperscript{9}, Pedro Teles\textsuperscript{10}, Sophie Jacob\textsuperscript{11}, Eleftheria Carionou\textsuperscript{12}, Panagiotis Askounis\textsuperscript{12}, Maria Grazia Andreassi\textsuperscript{13}, Danielle Berus\textsuperscript{4}, Isabelle Clairand\textsuperscript{11}, Jérémie Dabin\textsuperscript{2}, Jad Farah\textsuperscript{11}, Joanna Jurewicz\textsuperscript{8}, Renato Padovani\textsuperscript{13}, Danijela Arandjic\textsuperscript{1}

\textsuperscript{1}Vinca Institute for Nuclear Sciences, Serbia
\textsuperscript{2}Belgian Nuclear Research Centre, Belgium
\textsuperscript{3}Radiation and Nuclear Safety Authority, Finland
\textsuperscript{4}University of Brussels, Belgium
\textsuperscript{5}University Medical Center of the Johannes Gutenberg University Mainz, Germany
\textsuperscript{6}Universitätsklinikum Bonn - Rheinische Friedrich-Wilhelms-Universität, Germany
\textsuperscript{7}Norwegian Radiation Protection Authority, Norway
\textsuperscript{8}Nofer Institute of Occupational Medicine, Poland
\textsuperscript{9}Fédération des Hôpitaux Luxembourgeois, Luxembourg
\textsuperscript{10}Associação do Instituto Superior Técnico para a Investigação e Desenvolvimento, Portugal
\textsuperscript{11}Institute for Radiological Protection and Nuclear Safety, France
\textsuperscript{12}Greek Atomic Energy Commission, Greece
\textsuperscript{13}National Research Council – Institute of Clinical Physiology, Italy

Following the results of a number of studies on radiation cataractogenesis, the International Commission on Radiological Protection (ICRP) re-evaluated the dose limit for the eye lens, based on the new findings that, at relatively high exposures (>1 Gy), lens opacities may occur within a few years; however, at lower doses and dose rates, similar to those that might be encountered in occupational practice in medicine, visually disabling cataracts may occur over many years.
The European epidemiological study on radiation induced lens opacities among interventional cardiologists (EURALOC) was designed to provide better understanding of the relationship between radiation dose and risk of lens opacities, particularly at dose levels commonly occurring in occupationally exposed workers in interventional procedures. It will provide direct human data to evaluate dose-response and possible thresholds in realistic exposure conditions and with clinical assessment of lens opacities as the end-point.

Following idea of harmonised data collection, methodology of EURALOC includes: a) an extensive methodological development for dosimetry allowing the reliable dose estimates; b) evaluation of eye lens opacities; and c) quantitative risk estimation with precise and comparable data in a large cohort of exposed and non-exposed subjects.

The objective of this talk is to present methodological developments and results of eye lens dose assessment accomplished under EURALOC project, as well as to describe the process of recruitment of interventional cardiologists and unexposed persons, including the performance of eye examinations and classification of lens opacities for each recruited person. These results will be further used to determine a possible dose-response relationship in the group of interventional cardiologists.

Eye lens doses were assessed using two approaches: data from collected individual information on working practices using combined with eye lens dose values from literature and secondly, from the individual whole body dose values converting them to eye lens doses. In addition, eye lens dose measurements were performed in routine practice to validate both approaches and to determine their associated uncertainties. The obtained results revealed satisfactory agreement between measurements and retrospective dose assessment by two alternative approaches. No systematic errors have been found.

The recruitment phase of the EURALOC, performed according to the study protocol, included 399 exposed subjects working with interventional cardiology and a control group of 242 subjects. To maintain homogenous data structure, a joint database has been created and a pool of data was collected, including: data on other risk factors for lens opacities, work history data (occupational questionnaire on types and frequencies of interventional cardiology procedures performed by the exposed subjects which is essential for retrospective reconstruction of eye lens doses) and outcomes (results of the ophthalmological examinations, with scoring of lens opacities using the LOCS III system, separately for posterior subcapsular, cortical and nuclear opacities, as well as Scheimpflug images resulting in detailed and objective measures of lens translucency). The later activity was the most challenging, owing difficulties in providing a good quality images and subjectivity of the scoring system.

Preliminary results confirmed the feasibility of the study by successful formation of both exposed and non-exposed cohorts of subjects in the European collaborative EURALOC project. The study is well under way and the analysis is in progress.
ENEN+ Project - Attract, Retain and Develop New Nuclear Talents Beyond Academic Curricula

1C. Pesznyak, 2B. Bazargan-Sabet, 3A. Abdelouas, 4F. Tuomisto, 5M. Coeck, 6L. Cizelj, 7P D. Porras

1Budapest University of Technology and Economics (BME) Hungary
2Université de Lorraine (UL) France
3IMT Atlantique (IMTA), France
4Aalto University (AALTO) Finland
5Belgian Nuclear Research Centre (SCK•CEN) Belgium
6Jožef Stefan Institute (JSI) Slovenia
7European Nuclear Education Network (ENEN) EU/France

E-mail: pesznyak@reak.bme.hu


Nuclear technologies today exhibit unparalleled levels of safety and reliability. This has been made possible through considerable and long term efforts of the excellently educated and trained employees with outstanding safety culture in the industry, competent regulatory authorities, research, higher education and technical support (TSO) communities worldwide.

Early warning signs have started to emerge in the 90s in various European countries underlining the possible shortage of human resources and requirements for replacement of qualified nuclear personnel. The retirement of ageing workers, the lack of anticipation for preparing new generations of skilled workforce, negative public perception of nuclear and the lack of interest of young people to enter nuclear careers have been recognized as major difficulties encountered in just about all nuclear disciplines. This situation may give rise to the loss of nuclear knowledge, which might have already contributed to the reduced competitiveness of EU nuclear industry and could, in the future, also contribute to reduced safety and security of nuclear activities and installations.

The lack of new talents electing nuclear careers is closely linked to an early loss of interest in nuclear sciences and insufficient information about the nuclear careers available to both secondary school pupils and university students entering the Bachelor, Master of Science and PhD levels.

The primary motivation of the ENEN+ project is to substantially contribute to the revival of the interest of young generations in careers in the nuclear sector. This is to be achieved by pursuing the following main objectives:

- Attract new talents to careers in nuclear.
- Develop the attracted talents beyond academic curricula.
- Increase the retention of attracted talents in nuclear careers.
- Involve the nuclear stakeholders within the EU and beyond.
- Sustain the revived interest for nuclear careers.

The ENEN+ consortium will focus on the learners and careers in the following nuclear disciplines:

- Nuclear reactor engineering and safety.
• Waste management and geological disposal,
• Radiation protection and
• Medical applications.

For the ENEN+ project it is imperative to provide activities focused on the three main target groups of potential talents:

• **Secondary school pupils.** Attractive basic information on careers in nuclear will be developed, made available in national languages and complemented with an EU wide competition of pupils. A summer camp will be organized for the winners of the competition. Electronic tools, including social media, will be used as far as practical.

• **Bachelor students.** Most of the nuclear academic curricula within the ENEN association concentrate on master students. The existing efforts to attract bachelor students to pursue master education in nuclear will be strengthened by increasing the level of academic preparation for bachelor students. This may involve the reform of the pedagogy and culture of teaching in order to create exciting and engaging learning experiences, including opportunities for individual guidance towards nuclear careers and opportunities to interact with practitioners of nuclear.

• **Young professionals after graduation.** The nuclearization of graduates of non-nuclear sciences and technologies has been a considerable source of the nuclear talent throughout the nuclear era. Attracting more graduates to nuclearization may require strong support from the end-user and will be put in place through attractive e-information and opportunities for individual guidance towards nuclear careers coupled with opportunities to interact with practitioners of nuclear.

The list of participants can be found in Table1. The overall strategy of the ENEN+ work plan is to build a large collaborative effort integrating resources from university, research centres, industry and international organizations so that the common results obtained will deliver effective means to support education and training in 4 major nuclear fields (nuclear reactor engineering and safety, waste management and geological disposal, radiation protection and medical applications). The work packages and tasks can be found in Table2.

**Expected impacts:** In particular, the following results can be estimated to be a consequence of the proposed action:

• 2000 secondary school pupils reached by information;
• national secondary school pupil competitions in 15 EU member states;
• 60 secondary school pupils at the nuclear summer camp;
• 200 BSc students at 1 month nuclear internships;
• 100 MSc students at 2 month nuclear internships;
• 50 MSc students at 4 month EMSNE exchange visits;
• 20 BSc or MSc students presenting their work at the international conference;
• 50 PhD candidates or postdocs at 6 month exchange visits/internships;
• 20 PhD candidates or postdocs at 24 month research within EURATOM H2020 projects;
• 30 candidates for “nuclearization” at 1 month trainings;
• 35 candidates for “nuclearization” at 3 month trainings.

The ENEN+ budget is comprised by the following concepts: total costs: 3.2 € EC contribution: 2.9M€, mobility grants for students are 1.002.125€ (31%).
Table 1. List of participants

<table>
<thead>
<tr>
<th>No.</th>
<th>Participant organization name</th>
<th>Country</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>ENEN</td>
<td>EU/France</td>
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<tr>
<td>2</td>
<td>Université de Lorraine (UL)</td>
<td>France</td>
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<tr>
<td>3</td>
<td>Aalto University (AALTO)</td>
<td>Finland</td>
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<td>4</td>
<td>Belgian Nuclear Research Centre (SCK•CEN)</td>
<td>Belgium</td>
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<tr>
<td>5</td>
<td>Budapesti Műszaki és Gazdaságudományi Egyetem (BME)</td>
<td>Hungary</td>
</tr>
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<td>6</td>
<td>Universidad Nacional de Educación a Distancia (UNED)</td>
<td>Spain</td>
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<td>7</td>
<td>Centrum výzkumu Rež (CVR)</td>
<td>Czech Republic</td>
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<td>8</td>
<td>Jožef Stefan Institute (JSI)</td>
<td>Slovenia</td>
</tr>
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<td>9</td>
<td>Universidad Politécnica de Madrid (UPM)</td>
<td>Spain</td>
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<tr>
<td>10</td>
<td>Consorzio Interuniversitario per la Ricerca Tecnologica Nucleare (CIRITEN)</td>
<td>Italy</td>
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<td>Univerza vLjubljani (ULJ)</td>
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<td>Tecnatom S.A. (TEC)</td>
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<td>Universitatea Politehnica din Bucuresti (UPB)</td>
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<td>IMT Atlantique</td>
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<td>CEA Institut national des sciences et techniques nucléaires (INSTN)</td>
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<td>European Federation of Organisations for Medical Physics (EFOMP)</td>
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<td>21</td>
<td>Electricité de France (EDF)</td>
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<td>22</td>
<td>Nuclear GEN II and III association (NUGENIA)</td>
<td>EU/Belgium</td>
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</table>
Table 2. Work packages and tasks of the ENEN+ project

<table>
<thead>
<tr>
<th>WPs</th>
<th>Tasks</th>
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<tbody>
<tr>
<td>WP 1</td>
<td>Attract new nuclear talents in secondary schools</td>
</tr>
<tr>
<td>T1 1</td>
<td>Attractive E-information for secondary school pupils</td>
</tr>
<tr>
<td>T1 2</td>
<td>EU wide national nuclear competitions of secondary school pupils</td>
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<tr>
<td>T1 3</td>
<td>Summer camp for the finalists of the EU competition</td>
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<tr>
<td>T1 4</td>
<td>Explore the possibility of secondary school and university networking for nuclear matters</td>
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<tr>
<td>WP 2</td>
<td>Increase the attraction and retention of new talents among undergraduate students</td>
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<tr>
<td>T2 1</td>
<td>Strategies for attracting students at bachelor level</td>
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<tr>
<td>T2 2</td>
<td>Strategies for retaining students in nuclear fields</td>
</tr>
<tr>
<td>T2 3</td>
<td>Consolidate the EMSNE scheme and evolve it to EU Master of Science in Nuclear Disciplines</td>
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<td>WP 3</td>
<td>Attract and develop new talents through nuclearization</td>
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<tr>
<td>T3 1</td>
<td>Develop projections of workforce needs</td>
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<td>T3 2</td>
<td>Develop the ECVET based curricula for the identified profiles</td>
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<td>T3 3</td>
<td>Identification of best applicants from non-nuclear sectors</td>
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<td>WP 4</td>
<td>Develop new nuclear researchers beyond academic curricula</td>
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<td>T4 1</td>
<td>Enhance the experience and retention of PhD students and post-docs</td>
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<td>T4 2</td>
<td>Facilitate access to infrastructures and EURATOM projects</td>
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<td>T4 3</td>
<td>Consolidate and further develop the ENEN PhD event</td>
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<td>T4 4</td>
<td>ENEN Doctoral School on Nuclear Innovation</td>
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<td>WP 5</td>
<td>Consolidate and further develop European Fission Training Schemes and mobility</td>
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<td>T5 1</td>
<td>Consolidate vocational European Fission Training Schemes (EFTS) through voluntary accreditation</td>
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<td>T5 2</td>
<td>Coordinate the development of EFTS in niche applications</td>
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<td>T5 3</td>
<td>Propose sustainable ENEN+ grant mobility system</td>
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<tr>
<td>WP 6</td>
<td>Involvement of Stakeholders in EU and beyond</td>
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<tr>
<td>T6 1</td>
<td>Develop and implement communication strategy with industry and legislators</td>
</tr>
<tr>
<td>T6 2</td>
<td>Develop and propose the strategy for European nuclear ETKM</td>
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<tr>
<td>T6 3</td>
<td>Facilitate access to internships/apprenticeships</td>
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<tr>
<td>WP 7</td>
<td>Management and dissemination</td>
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<td>T7 1</td>
<td>Project management and QA</td>
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<td>T7 2</td>
<td>Dissemination</td>
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<tr>
<td>T7 3</td>
<td>Manage the mobility fund</td>
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<td>T7 4</td>
<td>Manage third parties &amp; subcontractors</td>
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NUCLEAR MEDICINE
COMMERCIAL PET DEVICES WITH LSO DETECTORS: WHETHER $^{176}$Lu ENTRIES PROBLEMS?

Antic Vojislav $^{1,2}$, Stankovic Koviljka$^2$

$^1$Center for Nuclear Medicine, University Clinical Centre of Serbia, Belgrade, Serbia, Pasterova 2, 11000 Belgrade, Serbia

$^2$School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11120 Belgrade, Serbia

Corresponding author: Antic Vojislav

e-mail: antic.vojislav@gmail.com

Introduction: The current PET devices contain scintillation crystals based on lutetium - LSO or LYSO. Lutetium $^{175}$Lu is incorporate with radioactive $^{176}$Lu (abundance 2.6%; $T_{1/2}=(3.56\pm0.07)\times10^{10}$ years; $\beta$- radiation ($E_{\text{max}}=596\text{keV}$), followed by three simultaneous $\gamma$-ray emissions (energies: 88, 202 and 307keV)). Taking into account long half-life, lutetium activity could be considered as constant - $52.61\pm0.36\text{Bq/g}$.

Objective: The objective of this paper is to estimate the amount of radioactive lutetium in PET-CT device from Clinical Centre of Serbia (with LSO detectors), and potential impact on the diagnostic information.

Methods: Based on the dimensions of the crystal, mass number and the total number of crystals, it was estimated the share of isotope $^{176}$Lu in the overall mass, and the associated radioactivity.

Results: According to the device technical specification, it was determined the detector size and calculated the mass of detector unit - 2,368g. Furthermore, there is 169 crystals per detector block and 144 detector blocks, which leads to the total detector mass - 57627g, from which the share of $^{176}$Lu is 1150g. Hence, there is ring distributed radioactivity of about 60.5kBq.

Conclusions: Natural radioactivity of lutetium poses no problem in standard clinical PET imaging, with excitation activities ~100 MBq and energy threshold of 350 keV, but:
1) Prevents the spread of energy window, without a significant compromise on the quality of the detected photons, mainly, due to the early rejection of the low energy photons and the low scatter factor

2) Could have influence especially at the end of dynamic studies using $^{11}$C or $^{15}$O

3) Affects on the QC examinations with low activities, with $^{68}$Ge point source (test tube activity ~ 5 kBq)

After replacement of the detector block or termination of device exploitation, the proper disposal of detector crystals is necessary.

Similar result could be obtained for detectors with LYSO crystals ($Lu_{2(1-x)}Y_{2x}SiO_5 :Ce$).
Tests for proton therapy treatment monitoring with in-beam PET: elemental composition analysis in space and time domains

Luca Brombal\textsuperscript{a,b,*}, Diego Barbosa\textsuperscript{c,d}, Nicola Belcaro\textsuperscript{c,d}, Maria G. Bisogni\textsuperscript{c,d}, Niccolò Camarlinghi\textsuperscript{c,d}, Luca Cristoforetti\textsuperscript{e}, Alberto Del Guerra\textsuperscript{c,d}, Francesco Fracchiolla\textsuperscript{e}, Matteo Morocchi\textsuperscript{c,d}, Slivia Muraro\textsuperscript{d}, Giancarlo Sportelli\textsuperscript{c,d}, Roberto Righetto\textsuperscript{e}, Marco Schwarz\textsuperscript{e,f}, Albana Topi\textsuperscript{g}, Valeria Rosso\textsuperscript{c,d}

\textsuperscript{a} Department of Physics, University of Trieste
\textsuperscript{b} INFN, Trieste, Italy
\textsuperscript{c} Department of Physics, University of Pisa
\textsuperscript{d} INFN, Pisa, Italy
\textsuperscript{e} Proton Therapy department, Trento Hospital, Trento, Italy
\textsuperscript{f} TIFPA INFN, Trento Italy
\textsuperscript{g} Department of Physical Sciences, Earth and Environment, University of Siena
\textsuperscript{*} Corresponding author.

Email address: luca.brombal@ts.infn.it

Introduction

Radiotherapy plays a major role in cancer therapy and roughly 50\% of the cancer patients receive this kind of treatment at some point [1].

Proton therapy is a leading edge radiotherapy technique which allows the delivery of a high conformal dose to the tumor minimizing the dose to the surrounding tissues. However, the steepness of the protons dose profile near the Bragg-peak makes this technique much more sensitive to spatial uncertainties than conventional photon treatments. Uncertainties in particle ranges, unexpected anatomical changes or patient setup errors may cause both over- and under-dosage in the target volume or increase the dose in adjacent normal tissues. For this reason, a treatment monitoring system would be highly desirable in clinical practice [2].

Since all the protons stop in the patient, the monitoring relies on the detection of secondary radiation: so far, Positron Emission Tomography (PET) is the most tested technique. In the case of proton therapy, the signal arises from the simultaneous detection of back-to-back photons (511keV) originating from \(e^+e^-\) annihilation, where the positron results from B+
radioisotopes produced in nuclear reactions between protons and the patient tissues. Since biological tissues are constituted mainly by oxygen and carbon, the main nuclear reactions of interest at clinical energies involve the production of $^{11}\text{C}$ and $^{15}\text{O}$ ($^{16}\text{O}(p,\,pn)^{15}\text{O},\,^{12}\text{C}(p,\,pn)^{11}\text{C}$) [3].

From the recorded signal, an activity distribution map can be obtained and useful information on the proton range can be inferred [4]. Furthermore, since the produced $\beta^+$ emitters are related to the tissue composition, the analysis of activity profiles and the time development of the recorded signal can provide insightful information on the elemental composition of irradiated objects.

In this study a planar in-beam PET prototype (DoPET) developed and built in Pisa was used and the measurements were performed at Trento Proton Therapy Center.

The aim of this work is to test the DoPET capability of providing information on the elemental composition of different target phantoms: some of the preliminary results are here presented.

**Materials and Methods**

The used system, i.e., DoPET, is a planar in-beam PET scanner composed by 2 $15\times15\ \text{cm}^2$ heads each consisting of 9 independent modules. One module is made of a $23\times23$ LYSO scintillator matrix ($\sim2\text{mm}$ pitch) coupled to an 8x8 multianode position sensitive photomultiplier Hamamatsu H8500. The read-out is performed by a custom front-end electronics connected to a FPGA, which embeds a coincidence processor with a time window of 3 ns. A more detailed description of the system can be found in [5].

The DoPET system is very compact and can be positioned directly on the treatment couch for data taking. Fig.1 shows the experimental setup at Trento Proton Therapy Center: the two planar heads are mounted onto an aluminum base 48 cm apart along with a phantom holder placed in the middle between the heads.

In this study 5 phantoms were irradiated: 3 of them are homogeneous slabs of Poly methyl metactrilate (PMMA), Polyethylene (PE), and brain equivalent tissue (BRAIN), while the other 2, referred to as PE zebra and BRAIN zebra, are composed by alternating 3 PMMA slabs either with 2 PE or 2 BRAIN slabs. The choice of the materials is based on their different content of carbon and oxygen as reported in Tab.1.

Each phantom was irradiated with a 130 MeV proton pencil beam ($\text{FWHM} \approx 10\ \text{mm}$) containing $10^{10}$ protons and the signal was recorded for 550 s after the beam was turned off (beam-off period).
The signal is collected in form of photon pairs detected within the energy window [350, 850] keV and the 3-D activity distribution is reconstructed via an ad-hoc developed Maximum-Likelihood-Expectation-Maximization algorithm.

Spatial distributions are reconstructed using data recorded within 120 s after the end of irradiation (an example is reported in Fig.2). From the 3-D distributions, 1-D profiles along the beam axis (referred to as z-axis) were extracted and analyzed.

When only time information is required, the signal, expressed as a coincidence rate as a function of time, recorded from the whole field of view is used: in this case also the subtraction of random coincidences, based on the delayed window technique, is applied.

Results

As it is shown in Fig.3, where the activity profiles of the two zebra phantoms are reported, different phantom compositions reflect into different profile shapes: the two activity defects due to the presence of PE slabs are more pronounced than those produced by BRAIN slabs. This behavior, given the integration time of 120 s, is mainly due to the absence of oxygen in Polyethylene. Furthermore, from both the profiles the correct thickness (20 mm) of the slabs can be estimated with an uncertainty smaller than 1 mm.

The difference in spatial profiles can be better understood observing the signal in the time domain. In Fig. 4 the coincidence rates as a function of time for the homogeneous PMMA and the two phantoms containing PE are reported. From the graph it is clear that the composition variation of the irradiated object strongly affects the time shape of the signal: within 300 s after the end of the irradiation the coincidence rate of the PMMA phantom is higher with respect to the phantoms containing PE. This fact is due to the higher percentage of oxygen in PMMA with respect to other phantoms. Moreover, after 300 s, the contribution of $^{15}$O has almost disappeared and only $^{11}$C is present, hence the signal of PE phantom is higher due to its higher content of $^{12}$C. The same considerations can be applied for Fig. 5 where BRAIN instead of PE material was considered: in this case, since the elemental difference between BRAIN and PMMA is smaller, the difference in the signal is less pronounced.

Discussion and References

With this study the monitoring capabilities of the PET monitoring prototype DoPET were
investigated focusing on the possibility to discriminate different material composition via both a space and time analysis. We have shown that, within a short time from the end of the irradiation (2 min), 1-D activity profiles allow the recognition of PMMA, PE and BRAIN slabs for two phantoms. We highlighted that the differences in the profile shapes are mainly due the different yield of $^{11}$C and $^{15}$O.

These measurements, performed with a simple irradiation geometry and high statistics of impinging protons, can be considered as a preliminary test to demonstrate how the time signal can be used for monitoring purposes. In the next future a quantitative estimation of each isotopic contribution as a function of the irradiated object elemental composition will be performed and additional materials, such as bone equivalent, will be studied.

Furthermore, experimental data, both in space and time domain, will be compared with the Monte Carlo code FLUKA. This kind of analysis may lead to interesting results in the context of clinical application: the comparison between the recorded time signal and a Monte Carlo simulation based patient model could provide a new tool for treatment monitoring.

On the other end, the capability to detect the presence of specific isotopes could be useful in some clinical scenarios. For instance, the irradiation of tissues containing calcium such as bones, are expected to produce short-lived potassium isotopes ($^{37}$K and $^{38}$K): detecting these $\beta^+$ emitters it would be possible to state if the irradiation of a bone structure took place or not.


Table 1: Density and chemical composition (fraction by weight) of the used materials

<table>
<thead>
<tr>
<th>Material</th>
<th>Density (g cm$^{-3}$)</th>
<th>H(%)</th>
<th>C(%)</th>
<th>O(%)</th>
<th>N(%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMMA</td>
<td>1.18</td>
<td>8.06</td>
<td>59.98</td>
<td>31.96</td>
<td>-</td>
</tr>
<tr>
<td>BRAIN</td>
<td>1.05</td>
<td>10.8</td>
<td>72.54</td>
<td>14.56</td>
<td>1.69</td>
</tr>
<tr>
<td>PE</td>
<td>0.94</td>
<td>14.4</td>
<td>85.6</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
Figure 1: Picture of the experimental setup mounted onto the treatment couch.

Figure 2: Lateral view of the PE zebra phantom activity distribution.
Figure 3: 1-D activity profiles. Comparison between the 2 zebra phantoms: z coordinate corresponds to the beam axis.

Figure 4: Comparison of the coincidence rate as a function of time for PMMA, PE and PE zebra irradiations. Some error bars are smaller than points.
Figure 5: Comparison of the coincidence rate as a function of time for PMMA, BRAIN and BRAIN zebra irradiations. Some error bars are smaller than points.
ANALYTICAL SOLVING OF BIOKINETICS MODEL FOR 90Y-DOTATOC

Marija Jeremić¹, Milovan Matović¹,³, Suzana Pantović³, Dragana Krstić², Dragoslav Nikezić²

1-Clinical Center Kragujevac, Centre of Nuclear Medicine,
2- University of Kragujevac, Faculty of Science,
3- University of Kragujevac, Faculty of Medical Sciences

Neuroendocrine tumors (NETs) are malignant tumors which most often attack gastrointestinal tract, pancreas, although they can appear in another organs like lungs, breast, thymus and urogenital system. Peptide receptor radionuclide therapies (PRRT) are successful therapeutic tools for treatment of patients with NET. This kind of therapy is based on usage of somatostatine analogues (DOTATATE, DOTATOC) labeled with strong beta emitters, such as ⁹⁰Y or ¹⁷⁷Lu. Based on beta particles energy, to treat larger tumors is recommended ⁹⁰Y, while ¹⁷⁷Lu is recomended for smaller tumors. Because of ⁹⁰Y is pure beta emitter it is difficult to estimate its distribution in organs after application. But, some facts are well known. After intravenous application, ⁹⁰Y labelled radiopeptide binds very quickly on tumor tissue, while the unbounded activity is excreted through the kidneys and urinary bladder. Certain amount of radioactivity is accumulated in kidney where can cause damage of kidney tissue and reduce kidney function. Due to this, it is of high importance to develop personalized dosimetric model for ⁹⁰Y in order to achieve better therapeutic efficacy of radiopeptides labelled with this radionuclide, as well as to achieve maximal reduction of kidney damage.

Our study included 14 patients who were treated by PRRT with ⁹⁰Y-DOTATOC in Centre for Nuclear Medicine, Clinical Center Kragujevac. All patients received between 2.7 and 5.4 GBq of ⁹⁰Y-DOTATOC. Blood samples were taken at the moment when the application of radiopharmaceutical was terminated, then each hour in first 6 hours and in interval of 6 and 12 hours, up to 72 hours after application of ⁹⁰Y-DOTATOC. According to the previously given instructions, all patients were collected and sampled urine of each urination, during whole period of 72 hours after application of ⁹⁰Y-DOTATOC. Measurements of ⁹⁰Y activity in blood and urine samples were done with liquid scintillation beta counter RackBeta, LKB Wallac.

In order to understand better biokinetics of ⁹⁰Y-DOTATOC in human body, two sets of differential equations were developed which described behavior of peptide in human body. Human body was considered to consist of five compartments and differential equations describe balance of ⁹⁰Y-DOTATOC in each of these compartments. Equations were solved analytically by software developed in house using Fortran 90 language. Output results are amount of ⁹⁰Y-DOTATOC in all compartments with the time during and after its application. Then, computed and experimental concentrations of ⁹⁰Y-DOTATOC in blood were compared for a given set of transfer coefficients λᵢⱼ. Final objective was to estimate these transfer
coefficients $\lambda_{ij}$ through the comparison of experimental and calculated values of $^{90}$Y-DOTATOC in blood and urine. Transfer coefficients were varied and computed activity in blood was compared with measured once in order to determine the best set of parameters. This will enable determination of absorbed dose in five organs of human body due to this therapy.
SVD analysis of backprojecting operator of ML-EM PET image reconstruction

Vencel Somai, Gábor Tolnai, Dávid Légrády
Institute of Nuclear Techniques, Budapest University of Technology and Economics
vencel.somai@gmail.com

Introduction

Maximum Likelihood - Expectation Maximization (ML-EM) is a stochastic iterative algorithm for positron emission tomography (PET) image reconstruction. Iterative methods offer improvements in resolution and stability over analytical approaches (e.g. the widely used Filtered-Backprojection), due to the more accurate modelling of the system, and the ability of accounting for noise structure. In exchange, their computational cost is tedious, so the image reconstruction is time-consuming even with nowadays’ average processing capacity. As the clinical and the pre-clinical use requires fast algorithms, it is indispensable to overcome the computational burden for allowing these kinds of reconstruction techniques to receive growing global acceptance in order to achieve better resolution in the course of both diagnostical and research use imaging. To mention only a few important examples, two of the key points in cancer treatment are the early detection and the recognition of spread cancer cells during therapy planning; both require resolution as high as possible. What is more, increased stability of a reconstruction algorithm allows decreased radiopharmacon dose, thanks to reduced sensibility to noisy data.

Materials and methods

The ML-EM method has two main steps: from an estimated source distribution we compute the detector response (forwardprojection) and by comparing it to the measurement data we improve our estimation (backprojection). These calculations use transport Monte Carlo simulations (usually parallelised on GPUs), and they are repeated in each iteration. The algorithm is extremely sensitive to the quality of the forwardprojection, so the corresponding physical modelling must be as accurate as possible. In contrast, the backprojection is a degree of freedom for a wide range of possible modifications, and indeed the modern reconstruction softwares make simplifications and neglect several physical effects in Monte Carlo simulations.

Our research started with the believed to be anomalous behaviour of GPU based Monte Carlo reconstruction code (PANNI – Pet Aimed Novel Nuclear Imager), developed in the Institute of Nuclear Techniques of BME. The novelty of the code was the ability of complete physical modelling in both (forward and back) projections in order to achieve better image quality, but the reconstruction happened to be better if one neglected most of the main effects (e.g. positron range, scattering, detector response) in the backprojection’s Monte Carlo particle transport simulations. The result seemed anomalous, because additional information of the system should have increased the quality of the reconstruction.

The main mathematical tool of our analysis was the singular value decomposition (SVD), as the backprojecting operator could be easily characterised by its singular values and vectors so the behaviour of the iteration scheme was easier to examine. As a result, we could point out that the speed of convergence of the algorithm strongly depended on the singular values and vectors of the backprojecting operator, and this was the main advantage of simplified
backprojection over full physical modelling. The increase in the speed of convergence was more significant than the missing information content about the system. However, the simplified operator does not mean an optimal form of backprojection, just easier Monte Carlo simulations. After having the missing knowledge about the influence of singular values and vectors to the convergence properties of the ML-EM algorithm, we made further manipulations in the backprojecting operator to amplify its advantageous effects. This modification was also implemented by means of SVD in the form of a posterior filtering operator, which we called SVD filtering.

**Results**

In noiseless test simulations, our SVD filtering method performed very well, and speeded up convergence with two orders of magnitude. However, it cannot be applied straightforward for the real, noisy case. A tomographical reconstruction process leads to Fredholm integral equations of the first kind. Unfortunately, the integral-operator of such a problem is accompanied by (spatially varying) blurring, so the inverse process is strongly sensitive to noise. As so, one must separate the information content form the measurement and modelling noise, which can be characterised (as a first approach) using Picard-condition. To summarise, the SVD filter mentioned above has to be modified and tailored for the amount of noise of a given reconstruction. After several simulations with varying noise content and source distribution, we obtained positive result, as SVD filtering performed better than the best backprojection setting so far with respect to resolution and the number of iterations needed for the reconstruction. We also found that the efficiency of the modification, i.e. the increase in convergence speed depended on the noise content of modelling, which also coincided with expectation.

**Discussion**

In conclusion, SVD filtering is an efficient improvement for ML-EM reconstruction technique and its significance increases parallelly with the available computational capacity. Furthermore, the implementation requires only a few lines of extra coding for a given reconstruction software.

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Comparison of activities of $^{131}$I to be administered to patients with benign thyroid disease according to one- and two uptake measurement protocols

Samac Jelena$^{1,3}$, Žeravica Radmila$^{2,3}$, Jakovljević Ana$^{2,3}$, Crnobrnja Veljko$^{2,3}$, Ilinčić Branislava$^{2,3}$, Vukmirović-Papuga Marija$^3$.

$^1$Department of Physics, Faculty of Sciences, University of Novi Sad
$^2$Faculty of Medicine, University of Novi Sad
$^3$Department of Nuclear Medicine, Clinical Center of Vojvodina

Corresponding author: Samac Jelena, samac.ki@gmail.com

Introduction: Radioiodine treatment for benign thyroid disease has been in use for over 60 years, and is considered to be well established. However, there is still a number of procedures in use, mainly differing in the aspect of the necessity of individual patient dosimetry [1]. Council Directive 2013/59/EUROATOM states that exposures of target volumes shall be individually planned for all medical exposure of patients for radiotherapeutic purposes [2]. Individual patient dosimetry, conducted through pre-therapeutic assessment of individual $^{131}$I kinetics in the target tissue based on administered tracer activity (TA) [3], is essential for optimizing the activity to be administered [4], contrary to fixed activity approach.

Materials and Methods: Effective half-life ($T_{\text{eff}}$) plays a major role in determination of activity to be administered to a patient. According to EANM SOP for pre-therapeutic dosimetry, if it is not possible to obtain a full set of data (5 radioiodine uptake - RIU measurements), the most accurate value of $T_{\text{eff}}$ is obtained with 2 RIU measurements (first 1-2 days, second 4-8 days post administration of TA) – protocol 1. Even thou, mostly used approach in clinical practice is measuring only one RIU (usually 24h post administration of TA), and estimating $T_{\text{eff}}$ to be 5.5 days – protocol 2. The goal of this study was to compare the calculated value of $T_{\text{eff}}$ with the estimated one, with the focus on it’s impact on calculated activity. Our study included 48 patients, in whom we obtained two RIU measurements, first at 24h and second at 96h post administration of TA. All patients were instructed to withdraw antithyroid drugs 4-5 days before administration of TA. Mass of the thyroid was determined by an ultrasound. Therapeutic dose to be delivered to the thyroid was prescribed by nuclear medicine physician, in accordance with clinical protocols [5]. We calculated activities following protocols 1 and 2.

Results: The mean effective half-life of $^{131}$I, measured during pre-therapeutic radioiodine testing, based on protocol 1, was found to be 7.51 days, ranging from 3.27 to 8 days. In 87.5% of cases, $T_{\text{eff}}$ was found to be between 5.5 and 8 days, caused by the difference between first and second RIU measurement of less than 33%. In the remaining cases (12.5%) the difference between two RIU measurements was more than 33%, resulting in $T_{\text{eff}}$ of less than 5.5 days. The mean difference in calculated activities to be administered, according to protocols 1 and 2, was 193 MBq, ranging from 11.5 to 1223 MBq.

Discussion: Higher the calculated $T_{\text{eff}}$ is, less activity needs to be administered for the same effect, since the thyroid retains it longer. As our results indicate, in 87.5% of patients, higher activity would be administered if we used only 24h RIU measurement, as compared to a set of two measurements. 2013/59/EUROATOM states that doses to non-target volumes and tissues shall be as low as reasonably achievable, justifying the need for the second RIU
measurement. On the other hand, economical and social factors need to be taken into account when requesting a patient to visit nuclear medicine department 3, instead of 2, times.

References:

OCCUPATIONAL EXPOSURE IN PET/CT DIAGNOSTICS: WHOLE BODY AND EXTREMITY DOSES

Antic Vojislav¹², Ciraj-Bjelac Olivera²³, Stankovic Jelena²³, Arandjic Danijela²³, Bozovic Predrag²³

¹Center for Nuclear Medicine, University Clinical Centre of Serbia, Belgrade, Serbia, Pasterova 2, 11000 Belgrade, Serbia
²School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11120 Belgrade, Serbia
³Vinča Institute of Nuclear Sciences, University of Belgrade, Mike Petrovica Alasa 12-14, 11001 Belgrade, Serbia

Corresponding autor: Antic Vojislav
e-mail: antic.vojislav@gmail.com

Introduction: In general, PET occupationally exposed staff receive higher doses compared to conventional nuclear medicine. Depending on the method of dose administration, facility design, staff capacity and workload, there are huge variations in radiation exposure.

Objective: The objective of this work was to estimate the amount of radioactive lutetium in PET-CT device from Clinical Centre of Serbia (with LSO detectors), and potential impact of ¹⁷⁶Lu natural radioactivity on the diagnostic information.

Materials and Methods: To improve the working practice in National PET Center, Clinical Centre of Serbia (semiautomatic dose administration), training was performed according to recommendations for the training of NM staff that are required to optimize the working practice (six technologists were involved). The activities included the use of nonradioactive material by the staff members in order to gain more experience and routine. Results were obtained based on the fourteen-month sample (seven plus seven, comparable number of patient studies: 681 versus 694). The improvement in reducing exposure time was considered, and consequently the doses obtained with electronic (ED) and thermoluminescence (TLD) dosimetry (dosimeters TLD-100, LiF: Mg, Ti / electronic dosimeters DMC 2000 X, MGP Instruments were used, for whole body dose measurements).

Results: The activities included the use of nonradioactive material by the staff members in order to gain more experience and routine. This action improved the efficiency and led to a time reduction up to 32% during the dispensing phase, 50% during the injection phase and nearly 40% during the removal of butterfly needle. Consequently, the staff doses become significantly lower. Normalised Hp(10) per unit activity per month for all six technologists dosimetry readings, both with ED and TLD presented in charts, presented this impressive difference.

Conclusion: ED and TLD dose analysis, both with ED and TLD, on the representative sample, indicates that radiation exposure of occupationally exposed nuclear medicine personell in great extent can decrease with suggested work optimization.

RADIOBIOLOGY
Preliminary results of the study of DNA damage in lymphocytes from patients undergoing prostate low dose rate (seed) brachytherapy.

Tímea Hülber\textsuperscript{1,2}, Zsuzsa S. Kocsis\textsuperscript{3}, Enikő Kis\textsuperscript{4}, Zsolt Jurányi\textsuperscript{3}, Géza Sáfrány\textsuperscript{4}, Csilla Pesznyák\textsuperscript{1,5}

\textsuperscript{1} Institute of Nuclear Techniques of Budapest University of Technology and Economics  
\textsuperscript{2} Radosys Ltd.  
\textsuperscript{3} National Institute of Oncology, Centre of Radiotherapy, Department of Radiobiology and Diagnostic Onco-Cyto-genetics  
\textsuperscript{4} National Public Health Centre - National Research Directorate for Radiobiology and Radiohygiene  
\textsuperscript{5} National Institute of Oncology, Centre of Radiotherapy  

Contact: hulber@reak.bme.hu

Introduction.

Two different types of biodosimetric assays were used in order to assess the biological damaging effect of low dose rate brachytherapy (BT) in healthy tissues in patients with prostate adenocarcinoma. Both chromosomal aberrations (CA) and micronuclei (MN) assays were proven by literature to be good indicators of received dose during partial-body radiotherapy [1,2,3]. This paper reports the preliminary results of a study evaluating the change in the number of chromosomal aberrations during BT in case of 5 patients treated by Iodine-125 seeds. Comparison of the manual scoring version of the two methods and a semi-automated MN scoring system was also done.

Materials and methods.

Samples: Five patients' blood was sampled just before the beginning of the therapy (No0) and at regular intervals during the treatment period (No1: 1 day, No2: 3 months, No3: 6 months, No4: 9 months and No5: 12 months after seed insertion). Twenty-four samples were examined in all.

Assays: In the course of the slide preparation cells were forced to start proliferation by phytohaemagglutinin, cultured in 10% 1640 RPMI and arrested in a special cell phase. In the case of CA test, cell cycle was stopped at metaphase with colcemid (= chromosomes are condensed, before being separated into two daughter chromatids), and in the case of MN test at anaphase with cytochalasin (= before the division of the cytoplasm and cell membrane occurs). The cells were fixed on slides and stained with Giemsa non-fluorescent DNA stain.

While the CA test can detect and classify each type of chromosomal and chromatid aberration (gap, deletion, isodeletion, chromosome breaks, dicentric chromosome, ring, translocation, exchange), MN assay proves only the existence of an aberration (in form of micronucleus...
appearing beside the two main nuclei) without indicating the precise type. Further downside of MN assay is not being specific to ionizing radiation; however, the so-called dicentric aberration identified during CA assay is a definite marker of irradiation. On the other hand, the objects that need to be detected during MN assay have simpler structure, which makes the scoring part of this method faster.

*Automation-aided MN scoring:* For the so-called semi-automated MN scoring we used Radometer-MN Series automatic microscope system developed by Radosys Ltd. The first step of this process is the automatic scanning the slide, scoring the binucleated cells (BN) and the MN. Then as a second step, using the system's dedicated software package we revised the BN and MN candidates proposed by the system and accepted or rejected them accordingly. In the case of low doses, this semi-automatisation as well could be a great advantage over manual scoring because examination of more cells even in the same amount of time can reduce the uncertainty of MN frequencies.

**Results**

We calculated four biodosimetric indices for each sample. 1-2) *Manual and semi-automated MN frequency* (= average MN number in BN cells normalized to 100 BN cells from the scoring of 200-500 BN cells) 3) *Frequency of dicentric (DIC) and ring (R) chromosomal aberrations together* (= number of DIC+R in 200 metaphases normalized to 100 cells). 4) *Total aberration frequency* (= Number of all type of aberrations in 100 cells).

In the case of samples prepared for MN scoring a huge variety of slide quantity was experienced, which influenced the efficiency of the automatic detection significantly. In order to decrease this variation, an extra cleaning cycle was added to the preparation protocol, which resulted in less MN artefacts but decreased the density of BN as well. This increased the manual scoring time significantly but left the time-requirement of automatic scoring unchanged. However due to the very low doses the signal to noise ratio was still really poor. Thus although the Radometer-MN Series automated microscope can conduct the whole scoring procedure automatically by itself, we had to heavily rely on the system's revision function.

**Discussion**

We can expect that the number of aberrations grows monotonously because during the low dose rate BT the seeds remain permanently inside the patient and give continuous exposure. Due to low doses, the statistical power of our small MN and CA values is relatively low. Figure 1. shows the semi-automatically determined MN frequency of the 5 patients versus elapsed time after seed insertion. Despite of large statistical uncertainties, slight increase in MN frequency can be seen here. That was confirmed with the calculation of an average value from the five patients' data, in spite of individual differences, the mean value shows a clear tendency (see in Figure 2). Since the inserted isotope's activity gradually decreases, what we can see here is probably the linear beginning of an exponentially saturating curve.
Based on Figure 3, we can conclude that there is a correlation between the amount of chromosomal aberrations and the number of micronuclei. Approximately one MN is formed while one CA appears. Our results are in good agreement with data of Wolff et al. [4] in case of rectal cancer patients. Prostate cancer patients have different weight and prostate volume, these factors have not been taken into account yet. Therefore the effect of these parameters on the results will be examined in the next phase of this study.

Taken together, the Radometer-MN Series automated microscope can conduct the whole scanning and scoring procedure automatically. The unique algorithm allows the identification and calculation of MN that are proportional to absorbed dose. Of course, revision of the software needs to be done regularly in order to increase its trustfulness.

References


Acknowledgment

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1. Figure: MN frequency and its uncertainty at different state of brachytherapy
2. Figure: Average of indices for the 5 patients
3. Figure: Comparison of semi-automated MN frequency and manually scored total aberration frequency
Biological dose estimation for different photon beam qualities used in radiation oncology

Gy. Farkas, Cs. Pesznyák, D. Béla, G. Székely, Zs. S. Kocsis, T. Major, Zs. Jurányi, C. Polgá尔

National Institute of Oncology, Centre of Radiotherapy, Budapest, Hungary

E-mail: pesznyak@reak.bme.hu

Introduction: The measurement of chromosome aberration frequencies in human lymphocytes is a valuable tool for assessing absorbed doses of ionising radiation in individuals [1]. Biological dosimetry has an important role in the investigations of radiation accidents, it can provide useful informations on therapeutical modalities as well. Although the effect of ionizing radiation in biological systems depends not only on the applied dose, but also on the used energy, however, such parameters like dose rate and filters are often neglected during radiation therapy [2]. Traditionally, radiation beam profiles are drawn to be “flat” because of dose computations and treatment planning.

The need for linear accelerators to produce “flat” radiation beam became unimportant after the introduction and successful implementation of IMRT technology. Removing the flattening filter (FFF mode) increases beam intensity, especially near the central axis. Increased intensity reduces treatment time, primarily for high-dose stereotactic radiotherapy/radiosurgery (SRT/SRS). However, experimental data of radiobiological effect of high dose rate is missing. Since FFF technology is relative new, beam characteristics need to be carefully studied prior to clinical use [3].

In our work peripheral blood lymphocytes were irradiated in vitro with a Varian TrueBeam linear accelerator and chromosomal aberrations were analysed. Samples were irradiated either with different energy levels or with different dose rates with flattening filter or FFF mode, then dose-reponse curves were compared.

Material and methods: Lymphocytes culture and metaphase spread preparation: Venous blood sample was obtained from healthy donors by venipuncture into Li-heparinized vacutainers. Samples were stored at 4 °C until the start of irradiation. 2-2 ml aliquots of blood were pipetted into cryotubes. Each tube was positioned in a plastic phantom filled with water in order to achieve homogenous dose distribution. Blood samples were placed in the isocenter of the radiation field and were irradiated at room temperature with different dose rates as 80, 300, 600 MU/min with 6, 10, 18 MV photon beams at doses between 0.5 and 8 Gy. Metaphases from lymphocyte cultures were prepared by standard cytogenetic techniques,
induced with phytohaemagglutinin M. Incubation time was 52 hours at 37 °C. Cell proliferation was inhibited with 0.1 ml colcemid (Gibco). Cell cultures were hypotonized with KCl, fixed, plated on glass slide, and stained with Giemsa. Minimum of 200 metaphases were scored and all aberration types were recorded: chromatid and chromosome fragments and exchanges, dicentrics, centric rings and translocations. Dose response relationships between the yield of dicentrics and photon sources were fitted by the linear-quadratic model. Dicentrics produced by single track events are proportional to the dose of radiation (αD), while the yields of dicentrics induced by two separate track events are proportional to the square of the dose (βD²). CABAS (Chromosomal Aberration Calculation Software) was used to fit the curves.

**Results:** Several factors are known to have an impact on calibration curves, such as differences between lymphocyte donors and culture protocols, slide preparation and scoring criteria. Therefore, to increase the accuracy of dose estimation, to avoid interindividual differences two donors’ blood was irradiated both with FF and FFF mode at 600 MU/ min 6 MV and 6 FFF-mode. More dicentrics plus centric rings were detected in FFF mode than in FF mode, and the difference was significant (At 6 Gy 219.50 ± 25.25/ 100 cells vs. 140.66 ± 16.75/100 cells, p.: 0.0171). Moreover, factors like the employed energy, and dose rate directly influenced the values of α and β. The value of β was higher at 6 MV (0.031-0.048 Gy⁻²) than at 10 MV (0.018-0.032 Gy⁻²). These data correlated well with previously published findings of other research groups [4, 5]. Calculations have revealed that between the 0-8 Gy dose range while the value of α decreases as the dose rate increased, the value of β increased parallel with the dose rate. The parameters α and β are highly negatively correlated.

**Discussion:** Higher aberration rate was found at low energy with high dose rate. Aberration rates were also higher at FFF mode compared to FF mode when larger fraction doses were used. These results might be important in the case of hypofractionated radiotherapy.

In this work we found that radiation photon energy has an influence on the shape of the dose–response curves; therefore, in the case of any accidental exposition the most appropriate dose curve should be used considering not only the type of radiation but also the energy and dose rate.

**References**


Physical and technical aspects of the determination of the dose to the patient in conventional diagnostic radiology

Nevena Ignjatov¹, Olivera Ciraj Bjelac²,³

¹Institute for Pulmonary Diseases of Vojvodina, Sremska Kamenica, Republic of Serbia
²Vinca Institute of Nuclear Sciences, University of Belgrade, Mike Petrovica Alasa 12-14, 11001 Belgrade, Serbia
³School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11120 Belgrade, Serbia

*Corresponding author: ignjatov.nevena@gmail.com

INTRODUCTION

This paper presents results of the patient dose assessment in conventional radiography in a large teaching hospital in Serbia with emphasis on physical and technical factors contributing to the patient exposure. Diagnostic radiology is the largest contributor to the population dose form man made sources of radiation and conventional radiography is considers to be the most frequent class of examinations.

MATERIALS AND METHODS

Dosimetric measurements for assessment of patient exposure were performed in the two radiography rooms at Philips Superix 1150 and Shimatzu x-ray machine. Total of 95 patients were enrolled in the study. The dose assessment was performed for the following types of diagnostic procedures:

- chest (PA);
- cervical spine (AP, LAT);
- lumbar spine (AP, LAT);
- pelvis (AP);
- knee (AP, LAT);
- upper extremities (AP, LAT).

Indirect method of assessment of patient dose was used. The entrance surface air kerma (ESAK) was calculated from the x-ray tube output and using actual exposure parameters for the particular patient, including correction for the geometry and BSF (Back Scatter Factor). Therefore, from measured X-ray tube output and using recorded tube voltage, focus-film-distance, tube current (mA) and the exposure time, ESAK was obtained using formula:

\[ \text{ESAK} = \frac{Y_o \cdot I \cdot t}{(L - (d + b))^2} \cdot \text{BSF} \]
Where \( Y_D \) is the x-ray tube output at distance \( D \) normalized by mAs (\( \mu \)Gy/mAs), \( I \) is the product of the tube current mA and the exposure time in seconds (s), \( L \) is the focus-film-distance, \( b \) distance from film and carrier of the patient, \( d \) is the patient thickness and BSF is the backscatter factor which depends on tube potential, device filtration and the size of radiation field.

For the purpose of calculating ESAK, X-ray tube output (air kerma) was measured using a calibrated semiconductor detector MPD Barracuda MPD (RTI Electronics AB, Goteborg, Sweden) in steps of 10 kVp in the range 50-120 kVp and at a distance of 1 meter from the X-ray tube focus.

For the indirect assessment of the effective dose, as a risk measure, it was used software package NRPB-SR262, with a library of conversion coefficients from ESAK to effective dose (Hart D, Jones D.G. Wall B.F Normalized Organ Doses for Medical X-ray Examinations Calculated Using Monte Carlo Techniques. Chilton, NRPB SR262, 1994) and tissue weighting factors from the International Commission of Radiological Protection document ICRP 103.

Uncertainty of dose assessment was performed for all radiographic examinations.

**RESULTS**

The typical exposure parameters for six radiography examinations are presented in Table 1, whereas results of dose assessment are presented in Table 2. The patient doses in terms of ESAK ranged from 0.07 to 20.3 mGy.

Table 1. Patient data and parameters of radiographic techniques for six radiographic procedures (9 projections)

<table>
<thead>
<tr>
<th>Radiography</th>
<th>Projection</th>
<th>Patient age</th>
<th>Patient weight (kg)</th>
<th>U(kVp)</th>
<th>I-t (mAs)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chest</td>
<td>PA</td>
<td>55 (20-79)</td>
<td>78 (55-110)</td>
<td>90 (50-112)</td>
<td>10*</td>
</tr>
<tr>
<td>Pelvis</td>
<td>AP</td>
<td>72 (56-80)</td>
<td>66 (53-80)</td>
<td>61 (55-65)</td>
<td>57 (56-60)</td>
</tr>
<tr>
<td>Cervical spine</td>
<td>AP</td>
<td>53 (42-65)</td>
<td>68 (63-72)</td>
<td>55(53-55)</td>
<td>23(22-25)</td>
</tr>
<tr>
<td></td>
<td>LAT</td>
<td>56.5 (32-77)</td>
<td>74.5 (62-94)</td>
<td>57(53-65)</td>
<td>23.5 (22-25)</td>
</tr>
<tr>
<td>Lumbar spine</td>
<td>AP</td>
<td>58 (45-80)</td>
<td>76 (58-93)</td>
<td>67.5 (63-81)</td>
<td>81 (80-90)</td>
</tr>
<tr>
<td></td>
<td>LAT</td>
<td>58 (45-80)</td>
<td>76 (58-93)</td>
<td>71 (64-77)</td>
<td>125 (124-126)</td>
</tr>
<tr>
<td>Knee</td>
<td>AP</td>
<td>66(41-73)</td>
<td>74(62-85)</td>
<td>64(54-70)</td>
<td>22(16-25)</td>
</tr>
<tr>
<td></td>
<td>LAT</td>
<td>62(42-72)</td>
<td>77(62-90)</td>
<td>58(44-65)</td>
<td>20(16-28)</td>
</tr>
<tr>
<td>Upper extremities</td>
<td>AP</td>
<td>62(30-76)</td>
<td>74(68-83)</td>
<td>42(40-49)</td>
<td>2(1.4-2)</td>
</tr>
</tbody>
</table>

*For Radiography of the chest were used only one value mAs
Table 2. Distribution individual ESAK values and mean values of effective dose for six radiographic procedures (9 projections)

<table>
<thead>
<tr>
<th>Radiography</th>
<th>Projection</th>
<th>Number of patients</th>
<th>ESAK (mGy)</th>
<th>Effective dose (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Min</td>
<td>Mean</td>
</tr>
<tr>
<td>Chest PA</td>
<td></td>
<td>29</td>
<td>0.5</td>
<td>1.6</td>
</tr>
<tr>
<td>Pelvis AP</td>
<td></td>
<td>5</td>
<td>4.2</td>
<td>5.6</td>
</tr>
<tr>
<td>Cervical spine AP</td>
<td></td>
<td>6</td>
<td>1.6</td>
<td>1.8</td>
</tr>
<tr>
<td></td>
<td>LAT</td>
<td>6</td>
<td>1.6</td>
<td>2.1</td>
</tr>
<tr>
<td>Lumbar spine AP</td>
<td></td>
<td>10</td>
<td>7.1</td>
<td>9.9</td>
</tr>
<tr>
<td></td>
<td>LAT</td>
<td>10</td>
<td>12.6</td>
<td>16.5</td>
</tr>
<tr>
<td>Knee AP</td>
<td></td>
<td>7</td>
<td>1.3</td>
<td>2.5</td>
</tr>
<tr>
<td></td>
<td>LAT</td>
<td>11</td>
<td>1.0</td>
<td>1.8</td>
</tr>
<tr>
<td>Upper extremities</td>
<td>AP</td>
<td>11</td>
<td>0.07</td>
<td>0.1</td>
</tr>
</tbody>
</table>

* The effective dose for Radiography of knees and hands is negligible, the upper limit for adults is 0.00017 mSv for hands and 0.001 mSv for knee and it makes no sense to calculate it for individual patient.

Measurement uncertainty of the type A is determined by the methods of statistical processing of results. Measurement uncertainty type A exists only if it is a measurement that is repeated several times. Result for measurement uncertainty of type A is:

\[ u_A = 4.5\% \]

The dominant part of the measurement uncertainty is the result of the measurement uncertainty type B, which is the result of the operation of a number of values influencing the measurement. In the analysis of the influence quantities are taken into account the causes of measurement uncertainty arising from geometry recording, instrumentation, X-ray apparatus and the patient itself.

Relative standard uncertainty of the type B (k=1): 15%

Combine the measurement uncertainty on the basis of the calculated value of the measurement uncertainty of type A and type B is:

\[ u_c = \sqrt{u_A^2 + u_B^2} = \sqrt{4.5^2 + 15^2} = 15.6\% \quad K=1 \]

For example the value of entrance surface air kerma for radiography of the chest can be written as:

\[ K_e = (1.6 \pm 0.16) \text{ mGy} \]
DISCUSSION

Systematic dosimetry measuring covers 95 adult patients in total, including both sexes and different body weights. Measuring defines ESAK reference values and assessed arithmetic means of effective dosage for six selected radiographic techniques and 9 adequate projections. The ratio of the maximum and minimum values for ESAK for different imaging techniques most varied in chest radiography in the PA projection. The amounts of dosage that patients receive are 1.6 mGy in average, which implies that the results are not in line with European diagnosing reference level of 0.3 mGy in PA projection for chest radiography, or DRL of 0.8 mGy in the case of Serbia. The chest radiography was made using the three phase Philips X-ray machine with manually selected exposure parameters. A major cause for the high doses is the value of tube current and time product fixed at 10 mAs as the chest radiography is usually recorded with 2-3 mAs. This exposition parameter is fixed on the radiography machine and there are no options for making changes. AEC option was available at Shimatsu x-ray machine and was used for all other radiographic imaging techniques in radiographic procedures. Lower levels of patient exposure were recorded at high frequency Shimatsu X-ray unit. The trend of counter-proportion between voltage and loading capacity of X-ray tube (shown in mAs) has been identified in all 6 radiography procedures. Irregularities shown above and significant variations of patient dosage during chest filming in PA projection would be avoided by removing these shortages.

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A Proposal for a Quality Control Protocol in Breast CT with Synchrotron Radiation

Anna Veronese, University of Trieste and INFN section of Trieste
Luca Brombal, University of Trieste and INFN section of Trieste
Adriano Contillo, University of Ferrara and INFN section of Ferrara
(corresponding author), contillo@fe.infn.it
Renata Longo, University of Trieste and INFN section of Trieste
Angelo Taibi, University of Ferrara and INFN section of Ferrara
Giuliana Tromba, Elettra – Sincrotrone Trieste S.C.p.A.
Fulvia Arfelli, University of Trieste and INFN section of Trieste

Introduction

Breast cancer is one of the most frequently diagnosed cancers and one of the leading causes of death for women worldwide. A key factor in treating and defeating such disease is early detection, which requires matching high image quality and an acceptable delivered dose. For these reasons, many research projects have been investigating 3D imaging modalities for the breast, in particular breast computed tomography (CT). Breast CT allows the complete volumetric reconstruction of the uncompressed organ. Even though the general concepts of breast CT have been known for many years, only recent technological advances allowed new research to bloom in the last decade. Many research groups all over the world are developing breast CT prototypes and starting clinical trials, following Boone’s paper in 2001 [1]. Among these, it is worth mentioning in particular Kalender’s group at the Friedrich-Alexander University Erlangen-Nürnberg (Germany) [2] and Ning’s group at the Rochester Medical Center (USA), which promoted a startup company to manufacture and commercialize their prototype [3]. In Trieste (Italy) the SYRMA-CT (SYnchrotron Radiation MAmmography-Computed Tomography) collaboration is setting up the first clinical trial of phase-contrast breast CT with synchrotron radiation (SR) at the SYRMEP (SYnchrotron Radiation for MEdical Physics) beamline of the Elettra synchrotron facility in Trieste [4]. In order to start a clinical trial it is necessary to develop and implement a complete quality control (QC) protocol. The aim of a QC protocol is to maintain the best image quality and at the same time to guarantee patient’s safety. We developed a proposal for a QC protocol for the SYRMA-CT set-up that comprehends a list of pre-patient checks, alignment checks, dose measurements and image quality measurements (alongside the associated measuring procedures). In developing the protocol we studied the existing ones for both mammography [5] and clinical computed tomography, taking into account also the protocol for the existing commercial...
breast CT system. All the conventional tests had to be reassessed for the specific SYRMA-CT experimental conditions and, if necessary, adapted to the experimental set-up in use. We also designed and built a custom made QC phantom for the Image Quality test.

Materials and methods

The radiation source of the SYRMEP Beamline at Elettra is one of the storage ring bending magnets of the synchrotron machine. The beam is monochromatic in the energy range 8.5 keV – 40 keV and the beam cross section in the patient examination room is about 220 mm (horizontal) × 3.4 mm (vertical). Besides planar mammography, performed by scanning vertically the imaged breast, the patient support was originally designed to perform breast CT as well, by rotating the breast in a pendant geometry outside of an ergonomically designed aperture at the rotation centre. The images are acquired with a CdTe single photon counting detector, PIXIRAD-8, placed at about 2 m from the organ to implement the so-called free propagation phase contrast technique. Regarding the above-mentioned QC phantom, its main structure consists of a water filled PMMA cylinder, 12 cm in diameter and 10 cm height, hung from the patient support through the aperture. The lower portion contains five rods of different materials of known attenuation coefficients, chosen to fit the range of materials composing the breast, in order to measure the system linearity and to calibrate the reconstruction. It also includes a PMMA insert with holes of different diameters to measure the low contrast resolution. A picture of the phantom is shown in Fig. 1.

Results

The tomographic images of the phantom were acquired at several beam energies within the range of interest, at different radiation doses and at two different heights, the first one in the lower portion of the phantom where the rods are placed, and the second one in the upper portion, that only contains water. The latter represents a uniform slice for evaluating the uniformity and the noise in the reconstructed images. We performed several image quality tests including visual check for artifacts and quantitative tests for CT numbers linearity, linear attenuation coefficients accuracy and low contrast resolution. The linearity of grey levels in the reconstructed image has been assessed through the image analysis of the two imaged portions. The QC phantom was also used to calibrate the system in terms of attenuation coefficients, by fitting the measured values against the theoretical linear attenuation coefficients for the specific energies. An example of such fit, performed at 38 keV with two different doses, is shown in Fig. 2.

Discussion

We designed and tested a QC phantom for the synchrotron radiation breast CT clinical set-up at ELETTRA, verified its compatibility with the experimental environment and performed the first quality control tests for SYRMA-CT project.
These preliminary measurements allowed to test the validity and feasibility of the proposed Quality Control protocol. Moreover, they provided useful indications on how to modify and improve the QC phantom prototype in terms of geometry and materials for a future design.

References


Fig. 1: The main structure of the QC phantom prototype: on the bottom the different material rods can be seen, as well as the “low contrast resolution” insert.
Fig. 2: Examples of the CT Numbers linearity results for a beam energy of 38 keV, at two different doses. The results show the fit of the grey level (in arbitrary units) versus the linear attenuation coefficient (in cm$^{-1}$).
Dose management and optimization in computerized tomography in Croatia - First results of the IAEA project

Ana Diklić¹, Doris Šegota¹, Goran Banušić¹, Petra Valković Zujić¹, Tomislav Benaković², Ivana Bjelobrk², Gordan Šarić², Ivana Kralik³, Krunoslav Marinčević⁴, Jelena Popić Ramač⁴, Neven Krivec⁵, Zoran Brnić⁵, Slaven Jurković¹,⁶, Dario Faj⁷

¹University Hospital Rijeka, Krešimirova 42, Rijeka, Croatia
²University Hospital Osijek, Josipa Huttlera 4, Osijek, Croatia
³State Office for Radiation Protection, Frankopanska 11, Zagreb, Croatia
⁴University Hospital Merkur, Zajčeva 19, Zagreb, Croatia
⁵University Hospital Sisters of Mercy, Vinogradska 29, Zagreb, Croatia
⁶Medical Faculty of Rijeka, University of Rijeka, Rijeka, Croatia
⁷Medical Faculty of Osijek, University of Osijek, Osijek, Croatia

Corresponding author: dariofaj@mefos.hr

Introduction

The number of CT (computerized tomography) scans performed is increasing rapidly throughout the past ten years and at this time CT contributes the majority of collective dose from diagnostic X-ray examinations in the world. Patient doses are usually higher than needed for setting up a diagnosis. Therefore, the optimization of CT procedures is necessary. For this purpose, an IAEA (International Atomic Energy Agency) project was initiated in Croatia in four major university hospitals.

Materials and methods

Four CT scanners were under the scope of the first data collection, three 16-slice and one 64-slice machines. After the validation of technical parameters of CT scanner used, an initial data collection was conducted prior to any optimization process. Local DRL (diagnostic reference level) values were obtained for five different clinical indications.

Results

Quality control measurement results were within the acceptability criteria. Local DRL values were mostly below the European DRL values with very few exceptions. Image quality scores showed mostly higher quality than needed for the indication.

Discussion
Results show that the optimization of the existing CT protocols is necessary in order to achieve acceptable image quality with least possible exposure. Furthermore, a team work of radiologist, radiation technologist and medical physicist is required to obtain better quality while reducing radiation risk.

References


2. Rehani MM. Limitations of diagnostic reference level (DRL) and introduction of acceptable quality dose (AQD). Br J Radiol 2015;88:20140344
Image quality evaluation of CT head protocols using visual grading characteristics (VGC) analysis

Francesca Pietrobon. 1, Cecilia Arrichiello 1, Chiara De Toffol 1, Nicola Zampieri 1, Paolo D’andrea 2, Matteo Mazzoli 2, Maurizio Amadei 2, Brunella Russoi 2, Matteo Costa 2, Marco Piuzzi 2

1 UOS Department of Medical Physics, S. Martino Hospital, Belluno, 32100 Italy
2 UOC Department of Diagnostic Radiology, AULSS n. 1 Dolomiti, Belluno, 32100 Italy

Corresponding author: Francesca Pietrobon, medical physicist

E-mail: francesca.pietrobon@ulss.belluno.it

Introduction

Radiologists are normally tasked with comparing image quality obtained using different setting or scanner. A correct balance between higher quality CT images and the radiation levels administrated is necessary. The quantification of radiation levels can be obtained using radiation dose quantities: CTDI vol (volume CT dose index) and DLP (dose length product). The performance of different image systems could be assessed by physical quantities as spatial resolution and low contrast detectability obtained from phantom images. To define effective and scientifically accepted methods of assessing clinical image quality, observer performance tests on images was proposed such as the European guidelines on quality criteria. Recently, the visual grading of characteristics analysis (VGC analysis) was proposed to measure image quality. The visual grading of the reviewers of two imaging techniques is described as the variation between the two situations, normal or abnormal, similar to a receiver-operator characteristic (ROC) study[1]. Since the most commonly CT examination requested and performed in the AULSS Dolomiti area is the CT head with or without contrast, software analysis of CT phantom and visual grading characteristics (VCG) analysis were used to compare different CT head protocols using on three different scanner.

Materials and methods

This research group had previously identified the locally CT head protocols performed in each participating scanner: a GE VCT 64-slice, a GE BrightSpeed and a Siemens Somaton Scope 16-slice CT scanner.

It had found five protocols: two VCT helical techniques (standard and AAPM routine Adult head[2]), two BrightSpeed techniques (axial and helical) and an axial Siemens Somaton technique.

To evaluate image quality, there are several methods: objective (phantom based) or subjective. A Catphan 600 phantom was scanned with these CT image systems using the
identified protocols. The images were analyzed by Iris QALite software and low contrast detectability and spatial resolution were estimated.

Five CT image data sets, one for each protocol, were reviewed by three local resident radiologists after data anonymization. They declared their assessment using a questionnaire grounded on the European Guidelines for this specific examination and a five-point scale from “confident that the criterion in not fulfilled” to “confident that the criterion is fulfilled”[3].

In this way, it has been obtained an absolute Visual Grading Analysis (VGA) that facilitate the quantification of subjective opinions. The VGC analysis provides the data from VGA studies and it is based on concepts developed for ROC analysis[4].

Even if VCG is less complex than ROC analysis, its statistical methodology, based a non-parametric rank-invariant approach, is appropriate to analyze data from an ordinal scale as this case.

The VGC-analysis was performed using the ROC analysis web-based calculator for ROC curves developed by J. Eng and R H Morgan[5].

The area under the ROC curve situated near 0,5 signifies that the image quality for the scan protocol on the vertical axis of the plot is identical to the image quality for the other scan protocol on the horizontal axis. The greater area indicates better image quality for the scan protocol on the vertical axis.

In this study, VGC analysis was performed to evaluate:

- Brightspeed axial protocol vs Brightspeed helical protocol;
- VCT helical standard protocol vs VCT helical optimized (AAPM) protocol;
- VCT helical optimized protocol vs Siemens axial protocol.

The areas under the ROC curve (\( \text{Area}_{\text{VGC}} \)), the standard deviation of the area and the resulting 95% confidence interval were showed in table 2.

**Results**

The radiation dose quantities, \( \text{CTDI}_{\text{vol}} \) and DLP, relative to the identified protocols are summarised in Table 1.

The areas under the ROC curve (\( \text{Area}_{\text{VGC}} \)), the standard deviation of the area and the resulting 95% confidence interval were showed in table 2.

Generally, higher quality CT images imply a higher radiation dose. CTDI values indicates that the axial protocol, in particular GE Brightspeed local axial protocol shows higher dose than other image system/protocol.
With regard to GE Brightspeed protocols, the low contrast detectability with axial protocol was better than helical protocol but the spatial resolution seems equal (Figure 1a and 1b).

These results are consistent with VGC curve (figure 3 and table 2) and it could be considered as a test to evaluate the significance of VGC method.

The comparison of Siemens axial and GE VCT AAPM helical protocols showed that Siemens appear to be greater than the low contrast detectability but GE VCT have a greater MTF. (figure 2a and 2b).

About VGC analysis, it can be stated that Siemens axial is statistically significantly better than GE VCT but with higher dose (figure 4 and table 2).

Finally, using VGC curve, GE VCT AAPM protocol produces radiological images better than GE VCT standard one with similar doses.

**Discussion**

The visual grading studies allow to use clinically available images without a gold standard during evaluation. VGC analysis is a valid statistical method for the data obtained by VGA experiments. In this study regarding CT head examination, dosimetry considerations and phantom results seems consistent with VCG curve.

In particular, there is a coherence between the low contrast detectability comparison and VCG curve one. This finding could be useful for other investigations.

**References**


Table 1. Radiation dose quantities for the locally CT head examinations

<table>
<thead>
<tr>
<th></th>
<th>CTDIvol [mGy]</th>
<th>DLP [mGycm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE VCT AAPM helical</td>
<td>37.00 ± 5.90</td>
<td>667.29 ± 108.58</td>
</tr>
<tr>
<td>GE VCT standard local helical</td>
<td>39.85 ± 3.52</td>
<td>704.52 ± 91.27</td>
</tr>
<tr>
<td>Siemens Somaton Scope standard local axial</td>
<td>46.76 ± 7.18</td>
<td>667.29 ± 108.58</td>
</tr>
<tr>
<td>GE Brightspeed standard local helical</td>
<td>38.29 ± 7.48</td>
<td>684.49 ± 129.40</td>
</tr>
<tr>
<td>GE Brightspeed standard local axial</td>
<td>65.68 ± 1.51</td>
<td>962.96 ± 59.81</td>
</tr>
</tbody>
</table>

Figure 1a Low contrast (GE Brightness axial/helical)

Figure 1b Spatial Resolution (GE Brightness axial/helical)
Figure 2a Low contrast (Siemens ax/VCT helical)

Figure 2b Spatial Resolution (Siemens ax/VCT helical)
Figure 3 VCG curve – Brightness hel/axial

Figure 4 VCG curve – Siemens ax/GE VGC
Figure 4 VCG curve – GE VCT AAPM/standard

Table 2 Summary of VGC curve

| Area \( \text{V} \) \( \text{GC} \) & Std.Dev(Area) (confidence level 95%) |
|-----------------|-------------------------------------|
| GE Brightspeed axial vs helical | 0.771 | 0.067 (0.637-0.905) |
| Siemens axial vs GE VCT | 0.790 | 0.0513 (0.682-0.8992) |
| GE VCT AAPM vs GE VCT standard | 0.820 | 0.0561 (0.7083-0.9327) |
Spectrometry measurement of scattered radiation in dental cone beam computed tomography

Jelena Stankovic Petrovic, Danijela Arandjic, Sandra Ceklic, Olivera Ciraj Bjelac, Predrag Marinkovic

1Vinca Institute of Nuclear Sciences, University of Belgrade, Mike Petrovica Alasa 12-14, 11001 Belgrade, Serbia
2Faculty of Physics, University of Belgrade, Studentski trg 12, 11000 Belgrade, Serbia
3School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11120 Belgrade, Serbia

*Corresponding author: jelena.stankovic@vinca.rs

Introduction

The three-dimensional (3D) cone beam computed tomography (CBCT) has expanded the field of oral and maxillofacial radiology due to relatively low dose and high spatial resolution. A CBCT system has a collimated x-ray tube that produces a cone-shaped beam, and a detector that can be produced as: (1) an image intensifier tube/charge-coupled device (IIT/CCD), or in recent years (2) a flat-panel imager with cesium-iodide and amorphous silicon. During the exam, the x-ray tube and the detector make one full (or partial) circular revolution around the patient, in that way producing a sequence of discrete two-dimensional (2D) images. A 3D volume is then formed by merging these 2D images together. The size of the CBCT field of view (FOV) is adjustable according to procedure and patient. It is reported that according to height of FOV, CBCT devices could be categorized as: dentoalveolar (<8 cm), maxillomandibular (8–15 cm), skeletal (15–21 cm), and head and neck (>21 cm). In addition to FOV, parameters as: x-ray spectrum, tube current and number of 2D projections are influencing delivered dose to patient and consequently ambient dose and dose to CBCT staff. The patient effective dose as high as 1073 µSv per CBCT procedure was reported [1]. Although new dental guidelines with focus on CBCT exams were published [2] by European Commission (EC), still there are a few published papers that express concerns about quality control, radiation protection in dental CBCT room, and lack of evidence-based data.

Scattered x-ray spectrum is also measured in CBCT room and reported in a few studies as a function of peak voltage of the tube (kVp), and measurement angle. The aim of this study was the spectrometry measurement of scattered radiation in a dental CBCT room as a function of FOV, image resolution and measurement angle. The results of the study are of importance in optimizing the radiation protection (RP) of medical staff and patients. In addition, the scattered energy distribution is of relevance for a personnel dosemeter calibration or for a RP measurement correction.

Materials and Methods

Measurements of scattered spectrum were performed around CBCT model SCANORA® 3Dx (Soredex, Finland) (Fig. 1). The CBCT system had an amorphous silicon (a-Si) flat-
panel detector. The x-ray tube was with fixed tungsten anode, focal spot size was 0.5 mm and anode angle was 15 degrees.

Cylinder PMMA phantom (16 cm in diameter and 15 cm in height) was placed instead of the patient, between the x-ray tube and the detector, simulating the patient head. The FOV of the system is adjustable according to the procedure and the patient size/age. In this paper the two FOVs settings and two resolution settings are considered, details are given in Table 1. The two surveyed resolution settings were: standard (SRES), and high (HRES).

The measurements were done by semiconductor spectrometer with cadmium telluride (CdTe) active volume, model X-123CdTe (Amptek, USA) shown in Fig. 1. The spectrometer was calibrated in Secondary Standard Laboratory at Vinca Institute of Nuclear Sciences. Calibration was done with calibration sources $^{241}$Am (59.5 keV), $^{133}$Ba (80.99 keV) and $^{137}$Cs (32 keV). An energy-channel conversion factor was 120 eV per channel. The detector showed a linear energy response with a resolution of 580 and 910 eV at 59.5 and 80.9 keV, respectively.

CdTe detectors are modern type of semiconductor detectors employed in photon spectroscopy. The X-123CdTe spectrometer is practical for clinical settings as it is compact and actively cooled with Peltier hybrid cooler. The detector, power supply, preamplifier, digital processor are packed in small casing (7 cm x 10 cm x 2.5 cm). The active volume of the detector is 3 mm x 3 mm x 1 mm, width, height and depth, respectively. Due to the relatively high atomic number (Cd: 48 and Te: 52), the CdTe detector with this volume size is efficient for detection of x-ray and gammas with energy lower than 100 keV. The lower detection limit is determined by 100 µm Be window. The peak tube voltage in this study was (90±5) kV, thus the study of scattered radiation in this x-ray tube vicinity is in compliance with spectrometer characteristics.

When using CdTe detector one should be aware that there is high possibility of escape peaks. This is due to relatively high energy of K-lines of Cd and Te: 26.7 keV and 31.8 keV, respectively, and small active volume. Additionally, the one of the most obvious characteristics of CdTe response above 50 keV is asymmetric peak tailing to the lower energy [3]. This is due to short lifetime of incident radiation induced holes, which leads to incomplete electrical charge collection.

In this survey, the spectrometer was placed d=1 m from the phantom central axis, perpendicular to the phantom longer axis, at the height of the phantom middle, as shown in Fig. 2. The scatter radiation variation with angle was also investigated. Measurements were performed for four angle values ($\phi$=120°, 200°, 280°, 320°, 0°) in one plane. The selected angles were conveniently chosen according to surveyed CBCT room characteristics.

**Results**

The results of scattered spectrum measurements and variations due to CBCT system settings and due to angle variation are shown in Fig. 3-7. The mean average energy for all measurement was 35.6 keV with standard deviation of 0.7 keV, minimum of 34.4 keV and maximum of 36.7 keV.

**Discussion**

The spectrum distortion due to escape peaks is clearly visible on Fig. 3. Probably, this is due the low quality of statistical information as the figure represents the spectra with the lowest intensity. The consequences of escape peaks are not severe in Fig. 4-7. The pile-up of counts is visible in all measurements, around 90 keV, but not in significant amounts.
As expected, the CBCT that had XM FOV and XRES resolution had the highest intensity. This was true for all surveyed angles. Subsequently, the XS FOV with XRES followed. The lowest spectrum intensity had the XS FOV and SRES exam setting. In average, the ratio between spectra intensities of XM FOV with SRES and the XS FOV with SRES was 2 for all surveyed angles. The ratio between intensities of spectra with SRES and XRES when FOV is small was 5. The biggest difference in spectrum intensity was found between XS FOV-SRES and XM FOV-HRES, and it was 8 and 20, for 0° and 120°, respectively. These findings are in agreement with reported results. The increase in the delivered dose with the increase of FOV and image resolution has been already reported [4]. Thus according to EC guidelines [2] and [5] it is suggested that CBCT equipment should offer a choice of volume sizes and examinations must use the smallest that is compatible with the clinical situation. The guidelines also suggested that if CBCT equipment offers a choice of resolution, the resolution compatible with adequate diagnosis and the lowest achievable dose should be used. Thus, the optimization of the exam settings to gain meaningful clinical information while keeping the exposure as low as reasonable applicable (ALARA) will lower patient and CBCT staff doses.

Spectrometry measurements provide a valuable description of the X-ray radiation in terms of intensity and energy distribution. In clinical conditions, the measurement of scattered radiation can lead to dose optimization of a patient and medical staff. Besides imaging device optimization, knowing scattered radiation field can provide crucial information for shielding calculation and dosemeter calibration. In this paper spectrometry measurement of scattered x-ray field in dental CBCT room is presented. The results showed that average energy of scattered radiation is near 36 keV. The other study finding is that CBCT settings like FOV and resolution had significant influence on scattered x-ray intensity, where larger FOV with the high resolution settings gave the highest intensities. This survey should be considered as introductory and these preliminary results will be employed as a basis in a future investigation.

This work was supported by the Ministry of Education and Science of Republic of Serbia (grant agreements III43009 and 451-01-967/2010-01).

References


Figure 1. CBCT SCANORA® 3Dx (left) and spectrometer X-123CdTe (right)

Table 1. CBCT settings

<table>
<thead>
<tr>
<th>Name</th>
<th>FOV Size [mm²]</th>
<th>Resolution Name</th>
<th>Voxel Size [mm]</th>
<th>Tube voltage [kV]</th>
<th>Electric current [mA]</th>
<th>Exposure time [s]</th>
</tr>
</thead>
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<tr>
<td>XS</td>
<td>50 × 100</td>
<td>SRES</td>
<td>0.4</td>
<td></td>
<td></td>
<td>2.3</td>
</tr>
<tr>
<td></td>
<td></td>
<td>HRES</td>
<td>0.2</td>
<td></td>
<td>90(±5)</td>
<td>10</td>
</tr>
<tr>
<td>XM</td>
<td>80 × 165</td>
<td>SRES</td>
<td>0.35</td>
<td></td>
<td></td>
<td>4.9</td>
</tr>
<tr>
<td></td>
<td></td>
<td>HRES</td>
<td>0.15</td>
<td></td>
<td></td>
<td>12.6</td>
</tr>
</tbody>
</table>

Figure 2. Geometry of the measurement with the starting positions of a-Si detectors and X-ray tube
Figure 3. Scattered spectrum, 0°

Figure 4. Scattered spectrum, 320°

Figure 5. Scattered spectrum, 280°

Figure 6. Scattered spectrum, 200°
Figure 7. Scattered spectrum, 120°
Quality Assurance of DEXA systems, necessity or overkill?

Ana Diklić¹, Doris Šegota¹, Slaven Jurković¹,²

¹University Hospital Rijeka, Rijeka, Croatia
²Medical Faculty of Rijeka, University of Rijeka, Rijeka, Croatia

Corresponding author: slaven.jurkovic@medri.uniri.hr

Introduction

Dual Energy X-ray Absorption (DEXA) scanning is currently the gold standard method for measuring bone mineral density (BMD). It is an effective parameter in assessing fracture risk, confirming a diagnosis of osteoporosis and monitoring the outcome of treatment.

By 2050, the worldwide incidence of hip fracture is projected to increase by 310% and 240% in men and women (Gulberg 1997), respectively. Consequently, there is a possibility of introducing population screening programs in EU member states.

The number of DEXA scans is rapidly increasing over the last two decades in both public and private medical institutions. Also, the technology of DEXA systems has led to higher exposure levels, shorter scan times, increased workload and increased shielding requirements. For example, patient dose from fan-beam is comparable to dental X-ray and approaching to dose of chest X-ray. Thus, a regular quality control including radiation protection and performance testing is required.

Materials and methods

Testing of DEXA scanners is obligatory by law in Croatia since 2009. However, the situation in practice is somewhat different. At University Hospital Rijeka DEXA machines are included in Quality Assurance program. Therefore, the protocol for independent evaluation of technical performance is developed in accordance to international recommendations.

The tendency for horizontal exchange of knowledge has motivated us to investigate the situation regarding DEXA practice on a national level. A survey was conducted among DEXA users. A questionnaire was distributed to all of them in order to obtain the basic information about their workload, quality control testing and personnel education.

Results

Results show the lack of quality control testing among Croatian DEXA users and also the lack of personnel education. Some further data investigation revealed actually even the lack of
understanding the terms of quality assurance/quality control. Quality control protocol and the results of this survey will be presented.

Discussion

Our intention is to emphasize the importance of incorporating DEXA machines and procedures in quality assurance programme. Nowadays, with the more frequent use of fan-beam and cone-beam scanners, the doses from bone densitometry are comparable to the ones from dental and chest PA X-ray examinations. Therefore, more attention must be given to this topic.

References


Establishment and implementation of QA / QC programme in diagnostic radiology in the west region of Croatia

Doris Šegota¹, Ana Diklić¹, Emina Grgurević–Dujmić², Vinka Kos³, Elvis Černeka⁴, Zdravko Jotanović⁵, Slaven Jurkovič⁶,⁷

¹University Hospital Rijeka, Rijeka, Croatia
²Public Health Center of Primorje and Gorski kotar, Rijeka, Croatia
³General Hospital Pula, Pula, Croatia
⁴Public Health Center of Istria, Pula, Croatia
⁵Orthopaedic Clinic Lovran, Lovran, Croatia
⁶Medical Faculty of Rijeka, University of Rijeka, Rijeka, Croatia

Corresponding author: slaven.jurkovic@medri.uniri.hr

Introduction

Quality Assurance programme on using ionizing radiation is mandatory in all EU member states. Even though terms of use of ionizing radiation for medical purposes are defined by the Croatian regulatory body since 2009 in diagnostic and interventional radiology this is still not implemented in most facilities. One of the main reasons is a lack of medical physicists in diagnostic radiology departments even at University hospitals. At this moment in Croatia, less than 5 medical physicists are involved in diagnostic and interventional radiology.

Materials and methods

A medical physicist at radiology department of University Hospital Rijeka is present since 2012 and Quality Assurance programme is implemented. As a result of the efforts of our Medical Physics Department for optimized, responsible and safe use of ionizing radiation in medicine at UH Rijeka, other health institutions in west Croatian region became strongly interested on developing and implementing their own QA programme. As they had no medical physicist and no ability to employ one, a collaboration between UH Rijeka and these institutions was initiated during 2015. It was agreed that medical physicists from UH Rijeka will periodically perform QC controls of higher complexity and provide education of technologists for daily and monthly QC tests. The next step would be the optimization of procedures. This cooperation include one general hospital, one special hospital and 2 public health institutions with 13 facilities.

Results
First results of QC tests showed a lot of problems with technical parameters of the equipment. Mainly, it was because of poorly maintained equipment. The most common problem was the functionality of the automatic exposure control in both, radiography and mammography units. Recommendations were given, some corrective actions have been made and some are still in progress. Next step of our collaboration is the optimization and establishing diagnostic referent levels. So far, the first step has been made which includes collecting data about radiographic practice and calculating entrance skin air kerma (ESAK) in radiography and average glandular dose (AGD) in mammography. First results showed a lot of variety between radiographic techniques and a wide range of patient doses for the same procedure in different facilities even within the same institution.

**Discussion**

Results showed that periodical QC is necessary as it is the only method for quantitative evaluation of the X-ray units. Additionally, reports and results from QC measurements upgraded the communication between users and service engineers. First results of the collected data about patient exposures showed that the optimization and also the additional education of technologists is necessary, especially since high patient doses were not generally linked to older equipment as it was expected. All of the mentioned activities lead to the conclusion that departments using ionizing radiation in their clinical practice have a strong benefit of the collaboration with a medical physicist. The main goal of quality assurance programme in diagnostic radiology, giving the adequate diagnostic information with least possible exposure, is now more achievable. For full QA implementation, including the optimization of radiographic practice, medical physicist is needed in radiology departments and therefore it should be a rule rather than the exception.

**References**

1. Zakon o radiološkoj i nuklearnoj sigurnosti, Pravilnik o uvjetima i mjerama zaštite od ionizirajućeg zračenja za obavljanje djelatnosti s električnim uređajima koji proizvode ionizirajuće zračenje (NN 41/13)


Quality and safety became very important issues in medical field. The process of digitalization and the development of the DICOM standard opened the possibility of tracking different parameters, which can be used as quality measures, patient dose being one of them.

Development of an application called ORQA (On-line Radiological Quality Assurance) started from the idea to establish a simple, not expensive and vendor independent tool, which will enable collection of different quality parameters from data available with digital images. ORQA is developed on open source platform, so there is no need to buy any licenses. Application is installed in user’s (hospital or other healthcare institution) information system and connected to PACS or to individual X-ray modality providing DICOM images. User selects parameters, which are to be collected and data is sent to central data warehouse where it is analysed and presented through web application. Custom reports can be produced using business intelligence software (Microsoft Power BI) and also presented via web application. Application can also collect phantom images, which can then be analysed automatically and data used for quality control (it is used in our mammography screening programme).

Primary aim of ORQA was to enable patient dose tracking but during its development we discovered that there are many other useful performance indicators within available data. ORQA can not only provide data for an effective quality assurance program but can also help in optimization of radiological procedures.

Some of the problems encountered during the development and set-up of the system will be presented together with examples on how the system can help in optimization of diagnostic procedures.
Introduction

Natural radiation sources give rise to the largest part of the annual effective dose for the general population. According to the UNSCEAR 2008 report, approximately 80% of the collective effective dose is due to the natural sources [1]. However, the number of medical procedures that include sources of radiation is increasing in many countries and the dose due to medical exposures is also increasing. Recently, medical exposures in USA became comparable with the dose due to natural sources and it is probable that in the near future the situation will be similar in other industrialized countries [1].

The increased number of medical procedures using radiation sources also causes the increased need for monitoring of occupationally exposed workers. TLDs have been traditionally used for this purpose, but active personal dosimeters provide a possibility to measure the dose rate in real time, which gives them an advantage over passive dosimeters. The usage of active dosimeters is increasing and in some countries they are mandatory for certain practices [2].

A question remains how suitable the modern active personal dosimeters are for use in the vicinity of different medical sources of ionizing radiation. While most or all of the dosimeters perform well in continuous X-ray or gamma-ray fields with the mean energy similar to the radiation emitted by isotope $^{60}$Co or $^{137}$Cs, many dosimeters have poor energy dependence and other problems might arise in pulsed radiation fields, mixed fields etc [2, 3, 4].
In this paper, a total of 13 active personal dosimeters of 9 different types were tested in continuous X-ray radiation fields with mean energies typical for diagnostic radiology. Special attention was dedicated to the low end of the X-ray spectrum.

**Materials and methods**

Testing was performed in Secondary Standard Dosimetry Laboratory of Vinca Institute of Nuclear Sciences. A Philips MG-320 X-ray unit was used to produce narrow series spectra, according to the specifications presented in [5]. The following qualities were used for testing: N-100 (mean energy of 83 keV), N-60 (48 keV) and N-40 (33 keV). All the tested dosimeters measure in terms of personal dose equivalent – $H_p(10)$. The reference values were determined by measuring the reference air kerma with a secondary standard and multiplying the air kerma with appropriate conversion factors available in [5]. Relative error for measurements in S-Cs radiation quality (662 keV) was included for comparison, because this is the reference quality for most active personal dosimeters.

For the purpose of this paper, relative error of dosimeter indication was evaluated according to the following equation:

$$E_r(\%) = \frac{M - H_p(10)}{H_p(10)} \times 100$$

where $E_r$ is the relative error of indication, $M$ is the active personal dosimeter indication and $H_p(10)$ is the reference value.

**Results**

The results given in Table 1 show that most of the dosimeters are unusable for the X-radiation with mean energy of 33 keV. Only 2 dosimeters have relative errors within ±30 %. More importantly, some dosimeters that according to manufacturers’ specifications can measure the X-rays of this energy, underestimate the dose for over 90 %, which can cause serious health risks for occupationally exposed workers.

According to manufacturers’ specifications, only 3 of the tested dosimeters are not suitable for the measurement of X-rays with mean energy of 48 keV. However, 2 of these dosimeters have relative errors smaller than 30 %. In total, 5 dosimeters underestimate the dose for more than 30 %, with maximum of 69 %.

When it is allowed for the uncertainty of the reference values, all dosimeters have relative errors smaller than 30 % for N-100 (83 keV) radiation quality. All dosimeters performed well in S-Cs radiation quality (662 keV), which is the reference quality specified by the manufacturer. The only exception is a dosimeter with different reference quality.
Discussion

The results of this research show that only a minority of active personal dosimeters is suitable for use in very low energy X-rays ($\leq 33$ keV) and that in some cases, manufacturers’ specifications are not reliable.

Many users from the field of diagnostic radiology perform dosimeter calibrations in only one radiation quality, which is usually S-Cs (662 keV), probably under assumption that if the dosimeter is working properly in the reference quality, it will be the case for the whole rated range. However, the results presented in Table 1 show that this is not true in almost half the cases and some manufacturers specifications do not correspond with the test results. The importance of checking this assumption increases for lower energies of the X-rays.

In order to obtain reliable results for measurements of personal dose equivalent in low energy X- or gamma-fields, it is necessary to use the equipment that is calibrated in the appropriate radiation quality, or is type tested. The problems presented in this paper can be only larger for applications of active personal dosimeters in the field of mammography, so additional research is needed.

The results of this research show that the type of detector is not the only factor influencing the suitability of dosimeters for low energy radiation. Same dosimeters based on both technologies show approximately 90\% under-response in N-40 radiation quality, while one dosimeter of each type had relative error smaller than 30\% in all the tests that were performed.

Special attention should be paid that this research didn’t include the behavior of dosimeters in pulsed fields, which could change the conclusions.

References


Table 1: Relative error of active personal dosimeters in different radiation qualities

<table>
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<th>Number</th>
<th>E_r (%)</th>
<th>Minimum rated energy (keV)</th>
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<td>N-40 (33 keV)</td>
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<td>N-100 (83 keV)</td>
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INTRODUCTION

The purpose of radiation shielding is to limit radiation exposures to employees and members of the public to an acceptable level. The objective of a shielding calculation is to determine the thickness of the barrier that is sufficient to reduce the air kerma in an occupied area to a value \( \leq \frac{P}{T} \), the weekly shielding design goal modified by the occupancy factor for the area to be shielded. Here is presented a method for determining the thickness requirements for barriers against scatter and leaking radiation in a radiographic room. The measurements were performed by RTI Barracuda equipment and R100B detector, which are very suitable for scattered radiation measurements because of the high sensitivity and the minimal energy dependence. Obtained thickness of the barriers was compared with calculated values according to NCRP report No. 147 which contains recommendation and technical information related to the design and installation of structural shielding for facilities that use X-rays for diagnostic imaging.

MATERIALS AND METHODS

Water equivalent plastic phantom is used as the radiation scattering material. Given RTG tube has Chest bucky module, so there are two primary barriers: floor and wall that RTG tube is facing to when chest scanning is performed. Measurements where performed with detector R100B connected in BiasW module on Barracuda cabinet and using RTI Ocean software (Fig 1.). Ocean software has different templates for measurements but in this case quick measurement template is used. Measurements where performed on two distances from the floor level: at 79 cm (which is the height of the machine table with phantom) and 163 cm, on each wall and each corner of the RTG room. Two different heights are used to check is there a difference between scatter and leakage radiation\(^1\). One set of three measurements where performed for each detector position and median value is taken. Settings of given RTG tube was: 81kVp, 80, mA, 4mAs, 50ms.
From calculating Transparency coefficients one can estimate barrier thickness (for different materials) from empiric figures that show dependence of transmission coefficient on barrier thickness\textsuperscript{[1]} Most common materials used are concrete, plate glass, lead (Fig 2.).

The weekly number of patients is 120, and the weekly number of patients taken for calculations of primary barrier for chest bucky is 60. Blueprint of the radiographic room where measurements are taken is given at Fig 3. Measurement setup is shown at Fig 4.

Calculations of the transparency coefficients for all given barriers are performed with all parameters and recommendations given in NCRP 147 report. Exceptions are that shielding goal is 0.02 mGy/week, and Occupancy factor is 1 for all rooms surrounding RTG room. Protection goal is to reduce the kerma to the background kerma level, in all shielded areas, even for the controlled areas.

**RESULTS**

In quick measurement setup Ocean software can measure exposure rate, exposure and time of the exposure. Direct measurement of exposure was not taken, since time of exposure was less than time set on the machine. To get real exposure value one must multiply exposure rate with set time on the machine (50 ms). Measured exposure is multiplied with the weekly number of patients to get weekly measured exposure rate. Transparency coefficient $b$ is calculated from\textsuperscript{[2]}:

$$ b = \frac{\text{shielding goal [mGy/week]}}{\text{measured exposure rate [mGy/week]}} $$

All measured exposure rates, their estimated weekly exposure rates, and estimated barrier thickness by measurements can be found in Table 1.

Calculating transparency coefficient require calculating air kerma at distance of the occupancy. For occupancy distance the recommended minimum of 0.3 m is added for every distance between RTG tube and barrier\textsuperscript{[1]}. Air kerma for secondary barriers is calculated from:

$$ K_{sec} = \frac{K_{sec}^1 \cdot N \cdot T}{d^2} $$

where: $K_{sec}^1$ is estimated air kerma per patient on the distance of 1 m from the tube combined for leakage and scatter radiation for all secondary barriers in RTG room for this type of procedure, and value of this factor is $4.9 \times 10^{-2}$ mGy/patient\textsuperscript{[1]}, $N$ is the number of patients per week, $T$ is occupancy factor which is 1 for all barriers, and $d$ is occupancy distance.

For primary barriers estimated air kerma calculations are similar:

$$ K_{prim} = \frac{K_{prim}^1 \cdot N \cdot T}{d^2} $$

$$ 104 $$
where: \(K_{prim}^{R}\) is estimated air kerma per patient on the distance of 1m from the tube for primary beam for this type of procedure, and value of this factor is 2.3 mGy/patient\(^1\). Note that estimated number of patients for this type of procedure is 60.

Transparency coefficient \(b\) is calculated from:

\[
b = \frac{\text{shielding goal [mGy/week]}}{\text{calculated air kerma [mGy/week]}}
\]

All calculated weekly air kermas, transparency coefficient and barrier thickness are shown in Table 1. where they are combined and compared with measured results.

**DISCUSSION**

Comparing transmission coefficient results obtained from calculations and results obtained from measuring with R100B detector, it is obvious that measured values are for one magnitude higher than the calculated values. Barrier thickness is growing exponentially with the reduce of the transmission coefficient\(^1\). Barrier thickness obtained in calculated method is four to six times greater than barrier thickness obtained from measured method. Knowing this, one can say that NCRP protocol is conservative method and fully according to ALARA principle. Note that occupancy factors are not included in the calculations, which would only increase transmission coefficient results obtained from both methods, and reduce barrier thickness. Note also that correction for occupancy distance was not taken in the measurements \((K_y=K_x(x/y)²)\) which would additionally lower measured air kerma rate with that increase transmission coefficient.

With air kerma measured at two heights from the floor level, there is a possibility to compare air kerma rate of leakage and scatter radiation. In NCRP report 147 estimated leakage radiation is greater than scatter radiation which can be confirmed from these results. At 163 cm from the floor level, detector is at the shorter distance from the RTG tube, and thus leakage radiation have greater influence on the outcome of the air kerma rate. At 79 cm from the floor level, which is the level of the phantom, scatter radiation have the main influence at the outcome of the air kerma rate.

**REFERENCES**

1. NCRP REPORT No. 147, Structural Shielding Design for Medical X-Ray Imaging Facilities, National Council on Radiation Protection and Measurements
TABLES AND FIGURES

Table 1. Combined results of measured and calculated barrier thickness

<table>
<thead>
<tr>
<th>Shielded area</th>
<th>Occupancy factor</th>
<th>Measured air kerma rate at 79 cm (µGy/s)</th>
<th>Measured daily kerma at 79 cm (µGy/s/week)</th>
<th>Transmiss. coeff. at 79 cm</th>
<th>Measured air kerma rate at 163 cm (µGy/s)</th>
<th>Measured daily kerma at 163 cm (µGy/s/week)</th>
<th>Transmiss. coeff. at 163 cm</th>
<th>Calculated kerma rate (µGy/s)</th>
<th>Transmiss. coeff. (calculated)</th>
<th>Propagation of shield thickness (primary barrier) (mm)</th>
<th>Propagation of shield thickness (secondary barrier) (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control room</td>
<td>1</td>
<td>13.65</td>
<td>81.99</td>
<td>0.24</td>
<td>16.82</td>
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<td>0.20</td>
<td>0.40</td>
<td>0.82</td>
<td>No primary barriers</td>
<td>Plate Glass (calc)=60 Lead (mess)=18 Concrete (calc)=60 Concrete (mess)=10</td>
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Fig.1. Ocean software
Fig. 2. Dependence of transmission coefficient on barrier thickness for different materials.

Fig. 3. Blueprint of the RTG room.

Fig. 4. Detector setup and measurement.
RADIOTherapy
Clinical implementation of „in vivo“ dosimetry with p-type EDP$^{3G}$ diodes

Kravic Ljubica, Rutonjski Laza, Petrovic Borislava, Baucal Milutin, Teodorovic Milan, Cudic Ozren, Basaric Borko, Golubovac Nemanja

Institute of oncology of Vojvodina, Sremska Kamenica, Serbia
ljubicasmiljanic86@gmail.com

Introduction: In vivo dosimetry represents a direct measurement of the dose delivered to the target volume in the radiotherapy. This specific dosimetry has been demonstrated to be valuable method for verifying dose delivery and has proved to be useful tool for quality assurance in radiotherapy. It is also a suitable method to both monitor the treatment delivery and to detect various errors early in the course of treatment. The accuracy in in-vivo measurements depends very much on the selected calibrations and corrections. A higher accuracy requires more time and effort in preparation. In this work was shown the determination of calibration and correction factors for p-type semiconductor diodes, for entrance in vivo dosimetry. The study was aimed to implement in vivo dosimetry as a part of quality assurance program in radiotherapy department. P-type EDP$^{3G}$ semiconductor diodes (IBA Dosimetry, Schwarzenbruck, Germany) for three energy ranges, 4-8 MV (Green), 6-12 MV (Red) and 10-20 MV (Yellow) were used in calibration and correction factors determination.

Materials and methods: Calibration and corrections factors for in vivo entrance dose measurements for four p-type EDP -10$^{3G}$ (green) semiconductor diodes for photon energy of 6 MV, four p-type EDP-15$^{3G}$ (red) semiconductor diodes for photon energy of 10 MV and one p-type EDP -20$^{3G}$ (yellow) for photon energy of 15 MV were determined as recommended by European Society for Radiotherapy and Oncology (ESTRO) Booklet No.5. All measurements were performed on the department’s two linear accelerators VERSA HD Delivery System (Elekta). Accelerators were calibrated to give 1 cGy/ MU at the depth of dose maximum for standard irradiation conditions (build up, SSD 100 cm, field size 10 cm x 10 cm) and all were equipped with record and verify (RV) system (Monaco/Mosaiq). Diode outputs were measured with the InViDos system (IBA Dosimetry, Schwarzenbruck, Germany). The software communicates with electrometer DPD-12 (emX; IBA Dosimetry, Schwarzenbruck, Germany). The overall factor for conversion diode reading to a measured entrance dose was obtained as the product of the dose calibration factor and all the correction factors for a particular beam.

Results: The procedure for all in vivo dosimeters was to set them up on a phantom surface. All p-type EDP$^{3G}$ semiconductor diodes for energy of 6, 10 and 15 MV were calibrated individually on the equipment on which they were to be used against a SSDL calibrated Farmer type ionization chamber FC65-G (IBA Dosimetry, Schwarzenbruck, Germany). The ionization chamber has been irradiated with the same treatment parameters at depth dose maximum. Set of correction factors accounting for non-reference conditions (for field size, SSD, wedge and gantry angle $C_F$) have been determined. (Table 1)

Discussion: Unexpected errors during treatment, may result in minor or even major deviation in dose delivered in comparison to the presribed and planned. One of the methods for significant improvement in treatment accuracy is shown to be in vivo dosimetry. In order to implement in vivo dosimetry with p-type semiconductor diodes in our deopartment, determinations of calibration and correction factors were done and the correction factors for every diode were in agreement with each other and also with correction factors reported in the literature. The next
step is performing entrance in vivo dose measurements on patients with aimed to get the full confidence that patients are being treated with the prescribed and planned dose.

References:


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Developing small beam dosimetry audits: a feasibility study

Tomislav Bokulic, Pavel Kazantsev, Paulina Wesolowska, Domonkos Szegedi, Luka Luketin, Joanna Izewska

International Atomic Energy Agency, Dosimetry and Medical Radiation Physics Section, Vienna, Austria.

Corresponding author e-mail address: T.Bokulic@iaea.org

Introduction

Accurately measured dosimetric characteristics of linear accelerators and well modelled dosimetry parameters in treatment planning systems (TPS) used for dose calculation are of ultimate importance for advanced radiotherapy techniques such as Intensity Modulated Radiation Therapy (IMRT) and different forms of Stereotactic Radiotherapy (SRT) and Stereotactic Radiosurgery (SRS). Achieving the desired dose distributions using small fields requires, among other things, accurate determination of small field output factors.

A quality audit feasibility study was designed at the IAEA Dosimetry Laboratory (DOL) to check high energy photon beam output factors and verify the agreement between the TPS modelled profiles and the profiles determined from the 2D dose distributions obtained from films.

Radiophotoluminescent (RPL) and optically stimulated luminescence (OSL) dosimetry systems, relevant for use in high energy photon dosimetry audits in radiotherapy, have recently been evaluated at DOL to determine their dosimetric characteristics and related correction factors [1].

This work presents the results of the small photon beam output measurements for a range of field sizes using RPLDs, OSLDs and a pinpoint ionization chamber (IC), as well as the results of a comparison of beam profiles obtained from the TPS and from the dose distributions measured with gafchromic films. The small beam output factors are furthermore compared to the reference dataset based on IROC Houston measurements in a number of radiotherapy centres [2]. The current study evaluates specially designed PMMA holders for dosimeters in terms of their robustness, reproducibility of positioning and ease of use.

Materials and Methods

The RPL dosimetry system used in this study consists of the glass rod GD-302M dosimeters (Asahi Glass Co.) and the Dose Ace FGD-1000 reader (Chiyoda Tech. Corp.). The RPLDs are silver activated phosphate glass rods. In the reader’s small aperture readout mode, the readout area located near the end of a dosimeter is 0.6 mm in diameter. The optically stimulated luminescent dosimetry (OSLD) system comprises NanoDot dosimeters and the microSTARii reader from Landauer. The NanoDot’s active carbon doped aluminium oxide (Al₂O₃:C) component is a 4.0 mm diameter by 0.2 mm thick disk. In order to validate the response of solid state dosimeters used within the study, measurements with the pinpoint IC (TW31014, φ 2 mm x 5 mm) were conducted.

Small field output factors were measured at a 10 cm depth and a 100 cm source-to-surface distance (SSD) for the following field sizes: 1×1, 2×2, 3×3, 4×4 and 6×6 cm² using the scanning
water phantom (MP3, PTW). Measurements were normalized to a 10×10 cm² field. For small field output factors determination the RPLDs, OSLDs and pinpoint IC were irradiated with 4 Gy, 1 Gy and 1 Gy, respectively. Five measurements were done per field. To ensure quick, accurate and reproducible positioning of the dosimeters in the scanning water phantom, a PMMA adapter for dosimeter holders was designed. The adapter fits to the Roos chamber holder; once it is aligned, the individual PMMA dosimeter holders (Fig.1) for the RPLD, OSLD, pinpoint IC and a film can be attached assuring central positioning at the beam axis. The alignment of the dosimeter in a holder with the central axis of the beam was done using the scanning water phantom. The holder correction factor that accounts for the different scattering and attenuation properties of PMMA and water was determined from the pinpoint IC measurements. As the small field point measurements were extended over several hours, they were corrected for the linac output variation that was monitored throughout the irradiation session.

All irradiations of the dosimeters were carried out with 6 MV photon beams from Elekta Versa HD, Agility MLC. The treatment planning system used for dose calculations was iPlan RT Dose (ver. 4.5.4) with XVMC algorithm. A 2 mm calculation grid was used and 1% statistical uncertainty of the dose in voxels was sought.

Small, 1x1 and 2x2 cm², field profiles modelled by the treatment planning system (TPS) were verified using gafchromic films. EBT3 gafchromic films irradiated with 4 Gy in a water phantom at 10 cm were used to determine the 2D dose distributions. The irradiated films were scanned with EPSON 11000XL scanner in 48 bit color depth, 150 dpi resolution and positive film mode with all image corrections disabled. The image analysis and the film profile extraction were done with Ashland FilmQA Pro software tools.

All point dose measurements were corrected for the volume averaging effects determined also from the 2D dose distributions recorded by EBT3 gafchromic films, as discussed earlier [3]. The pinpoint IC measurements were additionally corrected for beam perturbation related to non-water equivalence of the air cavity of the pinpoint IC [3].

Results

The summary of the measured small field output factors as a function of the field size, normalized to a 10×10 cm² field, is given in Fig. 2. The agreement within 1% between the TPS calculated and the IROC-Houston reference data, RPLD and OSLD measured data were obtained for field sizes larger than 1x1 cm². For 1x1 cm² field size the same agreement was within 2%. The output of the linac monitored before each of the small field output measurement session was constant and stable; the standard deviation of the output constancy check measurements for all sessions was 0.2%. The holder correction factor for 1x1 cm² was 0.993 and for other field sizes essentially equal to one.

Very good match of the profiles determined from gafchromic films and the TPS modelled profiles in both crossplane and in-plane directions were observed for 1x1 cm² and 2x2 cm² field sizes (Fig. 3). The profiles were quantitatively compared by the field size calculated at 20, 50 and 80 % relative dose and the largest observed discrepancy between the TPS modelled and the film measured profile was 0.6 mm.

The readout area of the RPLD is sufficiently small to fall within the flat part of the dose distribution represented by the in-plane and crossplane film profiles, requiring therefore no volume averaging correction. However, this is not the case for the cross section of the OSLD and
pinpoint IC. Therefore, RPLDs were chosen for the subsequent multicentre pilot testing of the audit methodology. Through the analysis of the 2D dose distributions obtained with films, the volume averaging corrections for fields 1x1 cm$^2$ and 2x2 cm$^2$ were found to be 1.036 and 1.010 for the pinpoint IC and 1.040 and 1.010 for OSLD.

**Discussion**

One of the steps in the development of dosimetry methodology for auditing more advanced technologies was the remote verification of TPS calculation of small field output factors of selected high energy photon fields in the range 2x2 cm$^2$ to 10x10 cm$^2$ [4]. Multicenter participants in this study obtained, on average, the agreement within 1% between the TPS calculated and the reference beam output factors data [2] for field sizes above 4×4 cm$^2$. For the field sizes less than 3×3 cm$^2$, TPSs used in this study overestimated the doses by 2% to 3% as compared to the reference data [4].

Recently characterized RPL and OSL dosimetry systems combined with newly designed dosimeter holders were successfully used in this feasibility quality audit study of small photon fields. Good agreement of small photon beam output factors measured for a range of field sizes with RPLDs, OSLDs and a pinpoint IC, with TPS calculated and reference small beam output factor dataset were found. The film profiles and TPS modeled profiles matched very well indicating that the TPS was accurately modelled. The dosimeter holders developed for this feasibility study were found practical, easily exchangeable and quick for reproducible positioning of a dosimeter in the scanning water phantom.

Based on these results a multicentre pilot audit study is being designed to fully test the methodology for small beam dosimetry quality audit. It includes the development of the pilot study instructions and data sheets, distribution of holders and dosimeters to participating institutions, evaluation of irradiated dosimeters, evaluation of profiles obtained from the TPS and from the film measured dose distributions, and establishment of the acceptance limits based on the results obtained.

**References**


Fig. 1. Holders (a-d) for RPLD, OSLD, pinpoint IC and gafchromic film embedded into the PMMA holder adapter.
Fig. 2. Small field output factors measured with RPLDs, OSLDs and pinpoint IC as a function of the square field side size. TPS data and IROC reference data are provided for comparison.
Initial experience with image guided and intensity modulated radiotherapy of gynecological cancer

Reka Kiraly, Csilla Pesznyak, Szilvia Varga, Nguyen A. Nhung, Gabor Stelczer, Istvan Szabolics Todor, Tibor Major, Csaba Polgar

National Institute of Oncology, Centre of Radiotherapy, Budapest, Hungary

e-mail: kiraly.reka@gmx.com

Introduction: The intensity modulated radiotherapy treatment of gynecological cancers is in an introduction phase at our institute. Our goal with the applied technology was to reduce the dose to the organs at risk, to determine the extent of the CTV-PTV margin and to check the patient setup accuracy.

Materials and methods: We contoured three different clinical target volumes (CTV) for 10 patients with gynecologic tumors: CTV1 includes the lymph nodes, CTV2 the parametrium and CTV3 the vaginal cuff. Then, we created the treatment volumes (PTV1, PTV2, PTV3) using the international recommendations of 7 mm, 10 mm and 15 mm margins. The patient's positional accuracy was checked with daily cone beam CT (CBCT), using the bony structures. Our IGRT protocol includes: "on-line" setup error identification and CBCT correction based on bony structures for the first three fractions. At the fourth fraction setup coordinates were adjusted by applying the systematic errors calculated from the average of the initial three measurements. From the fourth fraction, weekly check was performed and correction was made also when the error was more than 0.5 cm. Daily on-line correction was simulated by presuming a 3 mm residual error. CTV3-PTV3 extension was calculated with the "off-line" method by overlaying the planning and verification CBCT image sets. Two different radiation oncologists made the image matching independently. The CTV-PTV setup margin for CTV3 was calculated using the Van Herk formula. The treatment plans were created with the Varian Eclipse v11 planning system and the treatments were carried out with a Varian TrueBeam accelerator. The treatment was performed by using intensity modulated radiation therapy with rotating arc technique (RapidArc), with two full arcs. For all patients we applied the internationally recommended dose constraints on the target volume, bladder, rectum, small bowel and hip joints. The total dose ranged between 45 and 50.4 Gy with 1.8 Gy per fraction. Conformity number (CN) for PTV, V45 and V50 for organs at risk were used to assess and compare the treatment plans of RapidArc and 3D-CRT techniques.

Results: The margins between the CTV and PTV with or without IGRT were 1.0 cm vs. 1.5 cm, 0.9 cm vs. 1.3 cm and 0.6 cm vs. 0.7 cm in vertical, longitudinal and lateral directions for one oncologist, and 0.8 cm vs. 1.6 cm, 1.0 cm vs. 1.5 cm and 0.4 cm vs. 0.8 cm for the other radiation oncologist, respectively. According to the results of the image matching performed by
the two radiation oncologists, in case of daily on-line correction a 0.5 cm CTV3-PTV3 margin should be used. The target volumes dose coverage was satisfactory in all cases, but the RapidArc technology provided better dose conformity than the traditional 3D-CRT. The average conformity number CN values were 0.92 vs. 0.57; the RapidArc gave the better results. The dose constraints of the organs at risk for each patient were consistent with international recommendations. With the RapidArc technology the dose was lower than with 3D-CRT. The average values of V50 for the bladder were 50.5% and 16.4% and for the rectum 45.0% and 17.0% in the 3D-CRT and RapidArc techniques, respectively. For the small bowel the average values of V45 were of 30.5% and 15.9%.

**Discussion:** With the introduction of suitable contouring and IGRT protocols, the intensity modulated radiotherapy of gynecologic tumors can be done safely. To create the planning target volume different safety extensions should be applied for each clinical target volumes. The recommended 1.5 cm value for the vaginal cuff CTV-PTV margin seems to be sufficient even if no IGRT is performed. Use of smaller safety zone is possible only with more frequent verifications and on-line corrections.

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**Figure 1.** Determining the CTV3 extension using “off-line” image registration

**References:**


COMMISSIONING OF A PHOTON DOSE CALCULATION ALGORITHM BY COMPARISON WITH EXPERIMENTAL MEASUREMENTS

M.Sc Melisa Nurkovic1 Dr Paolo Maria Polloniato2 Dr Carlo Cavedon2
1Clinical Center of Montenegro
2Azienda Ospedaliera Universitaria Integrata-Verona-Italy
melisa.nurkovic@kccg.me

Introduction
The objective of this work was the commissioning of the Analytical Anisotropic Algorithm (AAA) photon dose calculation model implemented in the treatment planning system (TPS) Eclipse v.13.6.23 in Medical Physics department in Verona University hospital. The work was done by comparison with experimental measurements in homogeneous and inhomogeneous media, and testing two dimensional (2D) dose distribution in solid water phantom. The AAA was also tested against the present Pencil Beam Convolution (PBC) algorithm.

Material and methods
Configuration of the AAA: Beam configuration of the AAA model was done in the first part of this work putting the requested measured data into the TPS database for two Varian linear accelerators: Clinac600 for 6MV and DHX for 6 and 10MV. Since these linear accelerators were previously commissioned in Eclipse with the PBC model, there are no additional measurements required by vendor to configure the AAA algorithm [1]. The comparison between measured and calculated depth dose curves (PDDs) and beam profiles (PROs) at various depths for a number of different field sizes (FS) for open fields are done. The purpose of these checks, as recommended by IAEA TRS 430, is to confirm that these parameters in the calculation algorithm are correctly set, and the doses calculated with the AAA model agree with the data for the calculations to be used clinically [2].

Measurements in homogeneous water media were done on the linear accelerator Varian Clinac600. Set up for these measurements are given and describe in Table 1. Prior to these measurements the dosimetry in reference conditions was performed (Fref) using the IAEA TRS398 protocol [3], for the determination of absorbed dose in water in high energy photon beam. For the same set up point dose calculation were done in virtual water phantom with the AAA and the PBC using the Eclipse TPS v.13.6.23.

The measurements in inhomogeneous media were simulated by the CIRS Dynamic Thorax Phantom testing 2D dose distribution by means of PTW “2D Array”, a plan detector with 729 ion chambers. These were performed on the linear accelerator DHX using 6MV photon beam as well. Prior to the measurements with “2D Array”, the plan with field size 15x15 was created in the TPS. The tested plan was calculated with the same number of Monitor Units (200 MU) for AAA and for PBC. For 2D dose evaluation with a gamma criterion of 3% in dose and 3mm distance to agreement (DTA) VerySoft (PTW Freiburg, Germany) v.6.1 was used.
**Results**

Results of the comparison of the AAA calculated and measured PDDs and PROs for open fields beam configuration are given in Table 2 and Table 3. It provides an illustration on the accuracy that was obtained in PDDs and PROs calculation for different FS for 6MV.

*Measurements in homogeneous water media*

In Table 4 results from a point dose measurement in reference (F_{ref}) and non-reference conditions (F1-3), for the geometry and the same number of MU as it is described in Table 1 are shown. Measurement in reference condition was corrected for output of linear accelerator and compared with calculated dose with both algorithms AAA and PBC.

*Measurements in inhomogeneous lung phantom*

The results of testing 2D dose distribution in inhomogeneous lung phantom by means of the PTW “2D Array” plan detector are shown in Figures 1. The measurements for FS 15x15cm$^2$ was compared with the Eclipse calculation for both the AAA and the PBS algorithm. For evaluation of 2D dose distribution a gamma criterion of 3% in dose and 3 mm DTA is used, and the comparison is done with local dose and suppression dose below 10% of maximum dose of calculated volume.

**Discussion**

From the results of the comparison of the AAA calculated and measured PDDs and PROs, the dose difference is less than 1% for all FS at 10 and 20 cm depth, DTA is within 2mm at d_{max} and less than 1mm at 50% of dose for all field sizes except the 10x10 cm$^2$, where was found the maximum difference of 1.3 mm, Table 2. PDDs for square symmetric fields generally correspond well with the measurements. The AAA has a slight overestimation in the build-up region and in the area of maximum dose.

General consideration of calculated PROs is in well correspondence with measurements. For the field 4x4cm$^2$ and 10x10 cm$^2$ PROs calculated with the AAA have a small deviation compared to the measurements, and more in the upper-dose penumbra region for all depths, Table 4. For the larger fields PROs calculated with the AAA matches the measured profiles.

*Measurements in homogeneous water media*

The smallest deviation between measured and calculated dose is found for the measurement in reference condition (-0.6% calculated by AAA and -0.9% by PBC) as it was expected. For the asymmetric beam (F1) deviation between measured and calculated dose by AAA and PBC are found -3.6% and -3.5% respectively and for MLC field (F2) -3.6% and -3.9% respectively. Finally, two measurements with the oblique field (330°) and different position of detector along the central axis were measured (Filed 3a, 3b). The difference between the measured and calculated dose with both algorithms were less that 2.0%. Considering the difference between the dose calculated with AAA and PBC algorithm the differences were less than 1% for open beams (The last column in Table 4, $\Delta\%$).

*Measurements in inhomogeneous lung phantom*
The results for AAA at 6 MV with field size 15×15 cm\(^2\) were below the criterion of 3% in dose and 3 mm dTA, given that the only 72.7% of the evaluated points fulfilled the gamma criterion (figure 1). It is important to point out that the measurements were performed in conditions of electron disequilibrium, by adjusting the lung phantom directly on the surface of 2D array detector. The majority of the failed points were located in the proximity of interfaces between the lung and soft tissue (hot points from side of the lung, and cold from the side of the tissue), as well the hot points inside of lung and the cold points in the spinal cord. Area in the lung with more hot points is due to the missing of lung equivalent rod. For the PBC at the same conditions only 59.6% met the gamma criterion and the most failed points (cold points) were located in the soft tissue and in the spinal cord. It must be noted, however, that a full agreement is not expected for this test because of the experimental setup with air and lung portions of the phantom in close contact with the chamber array.

The obtained results allowed us to conclude that the AAA algorithm can be used in the clinical practice, and that it performs at least in an equivalent way compared to the previously used PBC. In selected situations, the AAA algorithm may be able to provide more accurate results as compared to experimental measurements.

References

Table 1. Set up for experimental measurements performed in reference (F_{ref}) and non-reference conditions in homogeneous media (F1-F5)

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<td>F3a</td>
<td>d=5cm</td>
<td>Oblique 330^0</td>
<td>10x10</td>
<td>90</td>
</tr>
<tr>
<td>F3b</td>
<td>d=10cm</td>
<td>Oblique 330^0</td>
<td>10x10</td>
<td>90</td>
</tr>
</tbody>
</table>

Table 2. Overview of the obtained accuracy of the calculated PDDs with the AAA represented by the dose difference at 10cm and 20cm depth and depth difference at d_{max} and 50% of dose for 6MV.

<table>
<thead>
<tr>
<th>PDDs</th>
<th>Open field</th>
<th>Open field</th>
<th>Open field</th>
<th>Open field</th>
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</thead>
<tbody>
<tr>
<td>Comparison parameters</td>
<td>4x4 cm^2</td>
<td>10x10 cm^2</td>
<td>20x20 cm^2</td>
<td>30x30 cm^2</td>
</tr>
<tr>
<td>ΔD10 (%)</td>
<td>0.4</td>
<td>0.4</td>
<td>0.4</td>
<td>0.1</td>
</tr>
<tr>
<td>ΔD20 (%)</td>
<td>0.1</td>
<td>0.3</td>
<td>0.2</td>
<td>0.0</td>
</tr>
<tr>
<td>DTA_{d_{max}} (mm)</td>
<td>2.0</td>
<td>1.0</td>
<td>0.5</td>
<td>0.0</td>
</tr>
<tr>
<td>DTA_{50} (mm)</td>
<td>0.9</td>
<td>1.3</td>
<td>0.9</td>
<td>-0.2</td>
</tr>
</tbody>
</table>

Table 3. Overview of the obtained accuracy of the calculated PROs by the AAA at depths: d_{max}, 5, 10 and 20 cm, represented by the distance to agreement at 80%, 50% and 20% dose level reflecting the quality of the penumbra modelling of 6 MV beam.

<table>
<thead>
<tr>
<th>PROs</th>
<th>Open field</th>
<th>Open field</th>
<th>Open field</th>
</tr>
</thead>
<tbody>
<tr>
<td>Comparison parameters</td>
<td>4x4 cm^2</td>
<td>10x10 cm^2</td>
<td>20x20 cm^2</td>
</tr>
<tr>
<td>d_{max}</td>
<td>DTA_{80} mm</td>
<td>-1.3/1.8</td>
<td>-1.0/1.6</td>
</tr>
</tbody>
</table>
Table 4. Results from point dose measurements performed according the set up described in Table 1. for reference ($F_{ref}$) and non-reference condition (F1-3) in homogeneous water phantom.

<table>
<thead>
<tr>
<th></th>
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<tbody>
<tr>
<td>$F_{ref}$ (open)</td>
<td>2.595</td>
<td>2.585</td>
<td>2.571</td>
<td>2.562</td>
<td>-0.6</td>
<td>-0.9</td>
<td>0.35</td>
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<tr>
<td>F1 (open)</td>
<td>3.213</td>
<td>3.201</td>
<td>3.090</td>
<td>3.093</td>
<td>-3.6</td>
<td>-3.5</td>
<td>-0.10</td>
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<tr>
<td>F2 (MLC)</td>
<td>2.636</td>
<td>2.626</td>
<td>2.534</td>
<td>2.527</td>
<td>-3.6</td>
<td>-3.9</td>
<td>0.30</td>
</tr>
<tr>
<td>F3a (oblique)</td>
<td>2.033</td>
<td>2.025</td>
<td>2.000</td>
<td>1.991</td>
<td>-1.3</td>
<td>-1.7</td>
<td>0.45</td>
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<tr>
<td>F3b (oblique)</td>
<td>1.502</td>
<td>1.496</td>
<td>1.483</td>
<td>1.482</td>
<td>-0.9</td>
<td>-0.9</td>
<td>0.07</td>
</tr>
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</table>
Figure 1. 2D dose distribution calculated with AAA (left), and PBC (right) FS 15x15cm² at 6MV. The gray area represent the region where gamma criteria were fulfilled and the red (hot) and the blue (cold) points are failed points.
Investigation of selected parameters of RPL dosimetry system in small aperture mode for the use in small field dosimetry

Domonkos Szegedi, Paulina Wesolowska, Tomislav Bokulic, Pavel Kazantsev, Joanna Izewska
International Atomic Energy Agency, Dosimetry and Medical Radiation Physics Section, Vienna, Austria
Corresponding author e-mail address: szegedi.domonkos@gmail.com, D.Szegedi@iaea.org

Introduction

The IAEA Dosimetry Laboratory (DOL) has been providing dosimetry audit service to numerous radiotherapy centres around the world for several decades. For this purpose, a thermoluminescent dosimetry (TLD) system has been used until recently. After investigating its complete dosimetric characteristics [1], a new radiophoto luminescent dosimetry (RPLD) system was introduced to the audit service.

To handle the challenges in small beam dosimetry, DOL develops an audit methodology for a range of small field sizes. For that purpose dosimeters with small active volume are used to measure the dose at a point thus avoiding the undesired volume averaging effect over a bigger area. RPL dosimeters can be read in the normal mode where the active volume is a 6 mm long part of the glass dosimeter. Another reading mode is the small aperture mode where the readout area is only a circle of 0.6 mm diameter, which requires no volume averaging correction even for 1x1 cm² field size [2]. This second mode is also called the high dose mode since it is suitable to measure doses from 1 Gy to 100 Gy.

When commissioning the RPL dosimeters for the small aperture mode one needs to determine the same dosimetric parameters and correction factors as for the normal mode (i.e. reproducibility, non-linearity of dose response, individual sensitivity of dosimeters, accumulated dose effect and energy dependence) [1, 3]. This work focuses on the determination of the sensitivity correction factors (SCFs) and the non-linearity of the dose response. In addition, the results of a self-test that was performed to check the system consistency including the correction factors used are presented.

Materials and Methods

The measurements were carried out with a Dose Ace FGD-1000 reader (Chiyoda Tech. Corp.) and GD-302M glass dosimeters (Asahi Techno Glass Corp.). Irradiations were performed in a solid water phantom with a holder specially designed for RPL dosimeters using the Co-60 units available at DOL. The absorbed dose to water measurements were performed before each RPLD irradiation session using the DOL secondary standard Farmer chambers connected to a Keithley 6517 electrometer. The RPL dosimeters are ready for use after annealing at 400 degrees for 20 minutes which erases the signal remained in the dosimeter. After the irradiation the glass dosimeters are preheated at 70 degrees for 30 minutes in order to stabilize the luminescent
centres and are cleaned in ethanol bath for 5 minutes to remove any dust particles from the surface of the dosimeters that could potentially change the readout signal. The RPLD dose can be calculated in the following way:

\[ D_{RPLD} = M \cdot N \cdot f_{SCF} \cdot f_{\text{lin}} \cdot f_{en} \cdot f_{\text{hol}} \]

where \( M \) is the readout signal, \( f_{SCF} \) is the sensitivity correction factor, \( f_{\text{lin}} \) is the correction factor for the non-linearity of the dose response, \( f_{en} \) is the energy correction and \( f_{\text{hol}} \) is the correction for the effect of the holder used for irradiation. \( N \) is the calibration coefficient of the RPLD system.

The SCFs were determined for 1000 dosimeters in the following way. 100 dosimeters were irradiated on the same day to the dose of 2 Gy in a solid water phantom. Then the sensitivity correction factor for the \( i^{\text{th}} \) RPL glass dosimeter was calculated as the ratio of the average corrected readout value of 100 dosimeters to the corrected readout value of a single dosimeter.

The readings of 100 dosimeters can take a few hours and the temperature can change in the meantime. The temperature dependence of the read out values was investigated previously in the laboratory and in this test the temperature corrections were applied.

\[ f_{SCF} = \frac{1}{100} \sum_{i=1}^{100} \frac{M_i \cdot f_{\text{temp,mag, irr},i}}{(M_i \cdot f_{\text{temp,mag, irr},i})}, \]

where \( f_{\text{temp,mag, irr},i} \) is the correction factor of the \( i^{\text{th}} \) dosimeter for the temperature, for position in the readout magazine and for the non-uniformity of the dose distribution during irradiation, since 20 dosimeters were irradiated at the same time in different positions around the Co-60 field centre at 5 cm depth using 95 cm SSD.

When irradiating the dosimeters to different doses the non-linearity of the dose response needs to be taken into account. The correction factor for this effect was calculated using the following formula:

\[ f_{\text{lin}} = \frac{M_{4\text{GY}}/D_{4\text{GY}}}{M/D}, \]

where the numerator is the RPL response per unit dose at 4 Gy and the denominator is the response per unit dose at a dose of interest. The reason for using 4 Gy as the reference point is to obtain a satisfactory signal to noise ratio since the readout area is small and therefore the dosimeter sensitivity is low. For the non-linearity test, RPL dosimeters were irradiated to 6 different doses in the range of 1-10 Gy with five dosimeters per dose.

Finally a self-test was carried out by irradiating 20 RPL dosimeters to 4 Gy dose. The RPLDs were split into 5 groups, each group containing a reference and three dosimeters considered the
end-user’s dosimeters. The dose measured ($D_{\text{meas}}$) with the user dosimeters was compared with the stated dose ($D_{\text{stated}}$) calculated from the irradiation time.

Results

The SCFs for 1000 dosimeters calculated from the readouts taken in RPLD reader’s small aperture mode are shown in Figure 1. Readings were repeated on two readers by three operators from the DOL and the average of the calculated SCF values was considered the final sensitivity correction. About 1% of the SCFs were removed from the analysis as they had outlying values that are not shown in the graph and after checking the dosimeters with a magnifying glass damaged edges or impurities were found. The final SCF values gave the standard deviation equal to $\sigma = 1.1\%$ with less than 5% of the results exceeding the $2\sigma = 2.2\%$ limits that are also indicated on the graph.

The RPL dosimeters showed a sub-linear response to the dose. The non-linearity correction was normalized to 4 Gy such that the correction for 4 Gy calculated from the linear fit of the data is equal to 1. The results are shown on Figure 2; the range of corrections is within 5% in the investigated dose range.

The results obtained in the self-test are presented in Figure 3. The ratios of the measured and stated doses were calculated from the readings performed with two different readers marked R1 and R2 respectively.

Discussion

The commissioning tests performed with the RPLD readers using small aperture mode showed that SCFs have to be applied for accurate small beam dosimetry with 95% of the correction factors within $2\sigma = 2.2\%$. The non-linearity of the dose response needs to be taken into account but only a small correction within ±1% has to be applied in the typical expected range of doses 4 Gy ± 2 Gy. The self-test results gave the average ratio of measured to stated doses ($D_{\text{meas}}/D_{\text{stated}}$) equal to 0.997 for 5 sets for both readers with the standard deviation of the results less than 1% (0.7% for R1 and 0.3% for R2). All dosimeter set results were within ±1% of the measured to stated doses. Similar results were reported for normal readout mode by Santos et al. [3]. In spite of the low sensitivity of the RPL dosimeters used in the small aperture mode, the results of the self-test were accurate showing very low scatter thus confirming that the RPLD system is suitable for small beam audit purposes.

References

Figures

![Fig. 1](image1.png)

Fig. 1. Sensitivity correction factor of 1000 glass dosimeters determined in small aperture mode. Dosimeters were irradiated to the dose of 2 Gy. 2σ limits are shown on the graph.

![Fig. 2](image2.png)

Fig. 2. The dose response non-linearity correction factor for doses ranging from 1 Gy to 10 Gy.
Fig. 3. The ratios of the measured and stated doses for 5 dosimeter sets. Readings were performed on two RPLD readers marked as R1 and R2. Error bars indicate the standard deviation of 3 dosimeters.
Evaluation of single and multiple isocenter frameless SRS plans for multiple brain metastases treatment

Attila Sarvari, Institute of Oncology Ljubljana, asarvari@onko-i.si

Introduction

At the Institute of Oncology Ljubljana we are using Brainlab frameless stereotactic radiosurgery (SRS) system. Treatment plan for brain metastases are prepared in the iPlan (Brainlab GmbH) treatment planning system and treated on Varian NovalisTX linear accelerator. In our study we have compared 10 patients with 3 or more brain metastases that were treated in the past two years. All patients were treated with plans that had multiple isocenters with doses varying from 16 Gy to 20 Gy depending on the size of the target volume. Additionally plans with a single isocenter were created in order to evaluate the benefit of the use of a single isocenter.

Materials and methods

The treatment preparation for the SRS starts with a CT scan with a slice thickness of 1mm. Additionally to the CT scan all the patients perform a MRI scan, with a slice thickness less or equal of 1mm, necessary for the contouring the organs at risk (OAR) as well as the target volumes. The clinical treatment plans consisted of multiple isocenters that were put in the middle of the planning target volume. The dose to the PTV was delivered in a single fraction using a low energy (6 MV) high dose rate (1000 MU/min) dynamic conformal arc (DCA) modality on a Varian NovalisTX linear accelerator. Every target was irradiated by 5 DCA beams spread by 30°-40° of table angle and around 100° of arc length. These parameters were depending upon the size, location and dose prescription of the target. The beam parameters (table angle, gantry start and stop angle, collimator) were defined in a way to avoid the irradiation of the critical organs as well as the other targets. In this way we have achieved the optimal dose coverage of the PTV as well as the lowest dose to the critical organs. In the study we included 10 patients with 3 or more PTVs but with the maximum number of 5. All 10 patients had a clinical plan done with the iPlan RT Dose 4.5.4 planning software and the dose was calculated by Monte Carlo algorithm (2.0 mm spatial resolution, 2.0% mean variance). In our study we made a plan with single isocenter that was put in the middle of all target volumes. The beam parameters (table angle, gantry start and stop angle, collimator) remained the same, because we wanted to compare doses that the organs at risk would receive. We compared only the most important organs at risk like brainstem, optic nerves, eyes and chiasm, since we used the dose volume prescription type to determine how the dose is calculated relative to the target structure. With this prescription type the TPS automatically calculates the best dose-volume coverage ratio, while taking defined constraints into account. This prescription type was used because it ensures us that a specific percentage of the planning target volume receives a specific percentage of the prescribed dose. These organs also had a large volume in order to avoid calculation problems. The evaluation is based on difference in dose volume histogram for every organ and as well as for the normal tissue. The difference is calculated in a way that we subtract the percentage of the organ volume at the certain dose in a plan with single isocenter from the plan with multiple isocenter (figure 1). For the PTVs we did not find any difference that would be worth to mention because of the prescription type used in TPS.
Results

We have averaged the results of all patients as well as all organs in order to get credible results (figure 2). The calculated results showed an increase of the dose to the organs at risk in case of a single isocenter. The difference are small (range of 5% for the normal tissue (figure 3)) and in the low dose region (below 4Gy). It has to be emphasized that the plans with a single isocenter did not have any beam parameter (table angle, gantry start and stop angle, collimator) optimization that could lead to a better sparing of the critical organs having the same dose coverage of the PTV.

Discussion

The results showed our prediction, that the dose to the critical organs as well as to the normal tissues will be higher when we plan with a single isocenter. If the plans with a single isocenter would be optimized (table angle, gantry start and stop angle, collimator) the difference would be even smaller. With this study we have made a first step of verification toward the use of a single isocenter in stereotactic radiosurgery treatment of multiple metastases.

References


Figure 1: difference in DVH for Brainstem for 10 patients and an average of all curves (curve AVG)
Figure 2: average difference in DVH for different type of OAR and average of all curves (curve Average)

Figure 3: difference in DVH for normal tissue and average of all curves (Average Tissue)
Fetal dose measurements in breast, head-and-neck and brain external beam radiotherapy of pregnant patients; a phantom study

Authors: 1Bencsik B, 1Stelczer G, 2Elek R, 2Fülöp N, 1Major T, 1Polgár Cs, 1Pesznyák Cs
1National Institute of Oncology
2National Research Institute for Radiobiology and Radiohygiene
E-mail: bara.bencsik@gmail.com

Introduction
In case the external beam radiotherapy of pregnant patients cannot be postponed post pregnancy, minimization of the fetal dose is essential. Depending on the tumour localization and the stage of gestation, in utero radiation exposure is associated with an increased risk of multiple severe complications. The aim of this study is to compare calculated dose by the treatment planning system (TPS) and measured dose in phantom.

Materials and methods
Nine treatment plans for breast, head-and-neck and brain tumour were made with Philips Pinnacle v3 TPS. The parameters of treatment plans can be found in Table 1 and Figure 1. Dose measurements were performed using Alderson RANDO® anthropomorphic phantom and the three different abdominal diameters were simulated with 3 (1st phase), 4 (2nd phase) and 5 (3rd phase) gel layers placed on the abdomen of the phantom. The electron density of the gel was close to soft tissue electron density with an average Hounsfield unit of 15. PTW 30001 Farmer ionisation chamber (IC) and natural Li thermoluminescent detectors (TLD) (Panasonic UT-807ATN) were used. In order to position the detectors accurately before every measurement session, CT image sets were created with metal markers on the TLD positions for each phase. The IC was placed at the same depth for each phase and tumour localization, while the TLDs were always positioned beneath the top gel layer in diamond shape around the IC. Three isocentres were determined on each CT set and treatment plans were developed according to the clinical practice. Adaptive Convolve dose calculation algorithm was applied. Breast and brain irradiation were scheduled with 2.67 Gy and 2 Gy dose per fraction, respectively, and planned with 3D conformal technique (3D-CRT). Head-and-neck irradiation was planned with step-and-shoot intensity modulated (SS IMRT) technique with 2 Gy fractional dose. Absorbed dose was calculated from the measured dose applying proper correction factors and compared with the calculated dose by the TPS.

Results
Measured and calculated dose depending on the distance between the detector and the edge of the irradiation field in longitudinal direction are summarized in Table 2. In case of breast irradiation, measured doses of the closest TLD varied between 1.88 cGy and 2.66 cGy depending on the diameter of the abdomen. Calculated doses at the same point were three times higher on average. At 12 cm distance from the field edge, ipsilateral and contralateral TLDs measured the average value of 0.93 cGy and 0.79 cGy dose, respectively. The measured average dose by the IC was 0.87 cGy for the different abdominal diameters, which is one-fifth of the calculated dose on average. In case of head-and-neck irradiation, the measured average dose by the closest TLD was 0.43 cGy for the abdominal phases and the calculated dose was one-third of the measured
value on average. In this localization, the IC measured average dose is almost seven times higher to the calculated average dose. For brain irradiation, the measured doses varied between 0.18 cGy and 0.08 cGy and calculated dose was maximum 0.15 cGy. According to absorbed dose of the closest detectors, the maximum fetal doses are 40 cGy, in case of breast irradiation (40.05 Gy total dose in 15 fractions), 12 cGy in case of head-and-neck irradiation (50 Gy total dose in 25 fractions) and 5.4 cGy brain irradiation (60 Gy in 30 fractions), respectively.

**Discussion**

The calculated dose by the TPS is not reliable far from the field edge so dose measurement in appropriate phantom is essential before the treatment, in order to estimate fetal dose correctly. Depending on the irradiation technique, the number and direction of the fields, measured dose is derived from scattering inside the phantom and from head leakage. The simulation of a real treatment needs proper shielding on the abdomen of the phantom to be able to estimate the dose from the scattering inside the body.

**References**


Table 1 Technical parameters of the treatment plans. (3D-CRT: 3D conformal radiotherapy; SS IMRT: step-and-shoot IMRT)

<table>
<thead>
<tr>
<th></th>
<th>Gantry [°]</th>
<th>Couch [°]</th>
<th>Collimator [°]</th>
<th>Wedge [°]</th>
<th>Energy [MV]</th>
<th>Technique</th>
<th>Field size (length x width)</th>
</tr>
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<td>15</td>
<td>0</td>
<td>90</td>
<td>15</td>
<td>20</td>
<td>6</td>
<td>3D-CRT</td>
</tr>
<tr>
<td></td>
<td>51</td>
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<tr>
<td></td>
<td>229</td>
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<td>6</td>
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<td>11.5 cm x 19 cm</td>
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<tr>
<td></td>
<td>229</td>
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<td></td>
<td></td>
<td></td>
<td>6</td>
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<td>head-and-neck</td>
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<td>6</td>
<td>SS IMRT</td>
<td>7 fields with 39 segments</td>
</tr>
<tr>
<td></td>
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<td>brain</td>
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Figure 1 Gel layer with the TLDs and the phantom measurement setup
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<thead>
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<th>BRAIN</th>
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<tbody>
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Establishing the Quantitative Gamma Evaluation Method as a standard IMRT QA verification procedure at the Oncology Institute of Vojvodina, Sremska Kamenica


**Introduction**: Since the IMRT (Intensity Modulated Radiation Therapy) planning was the next step compared to 3D-Conformal planning, new QA procedures needed to be found and implemented that were more suited to the complexity of IMRT treatment planning and dose delivery. Those QA procedures included a quantitative mathematical comparison of measured (reference) and calculated (compared) dose plain distributions through gamma evaluation method. This method uses the concept of dose difference (DD) and distance-to-agreement (DTA) first presented by Low et al. and today it is incorporated in most of the commercial gamma evaluation tools. The aim of this work was to establish neccessary IMRT QA verification procedures that use gamma evaluation algorithm, based on the results of measurements obtained at the Oncology Institute of Vojvodina in Sremska Kamenica.

**Materials and Methods**: Overall 59 plans (3D-CRT, IMRT and VMAT) were analyzed using gamma evaluation method as part of the MyQA Patients software (IBA Dosimetry). Dose distribution plains were recorded on the MatriXX detector (IBA Dosimetry) that was placed in a MultiCube solid water phantom (IBA Dosimetry). Each evaluated plan consisted of a measured (reference) and compared (TPS calculated) dose plain from the isocenter of MatriXX/MultiCube, gamma histogram and gamma result plain, as well as their statistical data.

**Results**: Dose plain distributions were analyzed on all plans using 3%/3mm global and local gamma evaluation tool in order to confirm passing rates ($\gamma \geq 95\%$ for global and $\gamma \geq 90\%$ for local gamma evaluation) and to establish the overall treshold value (20%). All plans were classified according to tumour localization and dose delivery method and presented accordingly with average passing rates and mean gamma values. Also, a 2%/2mm global gamma evaluation tool was used in IMRT (step and shoot) plans in order to establish a unique passing rate for our institution.

**Discussion**: Since the IMRT planning and dose delivery require very specific kind of TPS (algorithm) and linac, a certain number of possible random and systematic errors is attached to the issue. Gamma evaluation method is very important in uncovering these possible errors and represents standard tool in IMRT patient specific QA in any modern radiation therapy department today.

**References**:


Quality audit of IMRT technique dose delivery in Poland

Wojciech Bulski, Krzysztof Chelmiński, Wioletta Ślusarczyk-Kacprzyk, Piotr Ulkowski

The Maria Skłodowska-Curie Memorial Cancer Centre and Institute of Oncology,
Roentgen Str. 5, 02-781 Warsaw, Poland

Purpose:
The delivery of accurate intensity-modulated radiation therapy (IMRT) or stereotactic radiotherapy depends on a multitude of steps in the treatment delivery process. The purpose of this audit is to verify the dose delivery for an end-to-end clinical IMRT treatment executed with either a static gantry or VMAT technique. The extension of the programme to an end-to-end evaluation of advanced technology (IMRT) treatments provides an independent verification of the entire radiotherapy chain including imaging, the dose distribution calculated by the treatment planning system and treatment delivery. The methodology of the audit is presented here.

Methods:
The methodology of the end-to-end clinical IMRT audit was established within the framework of the CRP E2.40.16 project "Development of Quality Audits for Radiotherapy Dosimetry for Complex Treatment Techniques", run by the Health Section of the International Atomic Energy Agency (IAEA). A dedicated PMMA phantom was designed and manufactured. The phantom contains defined regions PTV (Planning Target Volume) and OAR (Organ At Risk). The phantom contains a special insert for placing radiochromic films and tubes with TLD powder. The participants of the audit are asked to perform CT scanning of the phantom, to prepare a IMRT treatment plan according to the given limitations concerning the homogeneity of the dose in the PTV, and limitation of the dose in the OAR, and finally to irradiate the phantom with films and TLDs inside according to the plan.

Results:
The audit in Poland is in the pilot phase. Until the end of 2016, 13 centres (out of 35) were audited. The results of film measurements in terms of percentage passing rate for gamma index evaluation (3 mm, 3% of isocenter dose) exceed 95% for 11 centers. The discrepancies in PTV and OAR between doses planned and determined with TLD were not higher than 5% in 10 and 9 centres respectively. The dose discrepancies higher than 5% require revealing and repeating of measurements. It is planned that all centres in Poland will be audited until the end of 2017.

Conclusions:
The audit was planned as a postal audit. However, for practical reasons it is carried out in the form of the visits to particular centres. Such form of the audit makes it possible to supervise the local staff in their and assure that the procedures are carried out correctly. A high impact of positioning errors on the results was observed. The results obtained with films are correlated with TLD. Already in this phase it may be stated that the elaborated methodology functions well in practice and makes it possible to evaluate the radiotherapy procedures in particular centres. However the minor improvements were needed.

Corresponding author: w.bulski@zfm.coi.pl
Comparison of Co-60 and linac based radiotherapy for DIFFERENT treatment sites

Msc. Rodina Myrku\textsuperscript{a}, Msc. Ortesa Dhima\textsuperscript{a}, Msc. Uada Bitri\textsuperscript{a}, PhD. Eduard Gershkevitsh \textsuperscript{b}

\textsuperscript{a} University Hospital “Mother Theresa“, Oncology Service, Tirana, Albania

\textsuperscript{b} North Estonia Medical Centre

Main author adress E-mail: rodina_cela@yahoo.com

\textbf{Introduction:} In this study, we perform a DVH comparative analysis for conformal radiotherapy plans using Cobalt – 60 unit (Best Medical Equinox 100 with customized blocks) and linear accelerator (Elekta Synergy platform with 80 MLC), by analyzing dose distribution in target volumes and organs at risk. The aim of the study was to determine which patients would benefit most from the Linac based delivery.

\textbf{Materials and Methods:} Thirteen patient’s plans that were previously planned with curative intend on Elekta XiO\textsuperscript{®} treatment planning system (TPS) for Co-60 unit were chosen. The plans were re-planned for Linac, a new machine available in the clinic. PTV coverage dose and dose volume histograms (DVH) for organ at risk were analyzed for both units. In addition to DVH other metrics such as conformity index (CI), homogeneity index (HI) and respective doses (min, max, mean) for PTV and organs at risk were compared. OAR data was also compared to QUANTEC data.

\textbf{Results:} DVH comparison of thirteen different sites using Cobalt and Linac beams showed that the use of MLC and different energies combination in Linac, resulted in reduced normal tissue dose, while the coverage of 95% of PTV have better results, respectively, average for all study patients with Cobalt is 81,78% and with Linac is 91,3%. According to the HI index, average values for Cobalt is 1,08 and Linac is 1,06. Lower value of CI for organs at risk was observed demonstrating better sparing of critical organs using Linac.

\textbf{Discussion:} The use of energy combination, MLC and additional small fields, gives better conformal plans and dose distribution for all cases using Linac machine rather than using Cobalt unit

\textbf{References:}

Dose of different image guidance techniques in breast radiotherapy

Gabor Stelczer1,3, DoraTatai-Szabo1, Jeno Palvolgyi2, Tibor Major1, Csaba Polgar1, Csilla Pesznyak1,3

1National Institute of Oncology, Budapest, Hungary
2Petz Aladár County Teaching Hospital, Győr, Hungary
3Budapest University of Technology and Economics, Hungary

E-mail: tataiszabodora@gmail.com

Introduction

Image-guided radiation therapy (IGRT) is a widespread technique to reduce set up errors and to improve the accuracy of patient positioning by using frequent imaging during the treatment course. The aim of this study was to make point dose measurements in a thorax phantom to compare the dose of six different image guidance techniques of three different linear accelerators.

Methods and materials

Measurements were performed on Elekta Synergy (E), Varian TrueBeam (V) and Siemens Artiste (S) machines. These linear accelerators offer different options to do image guidance. In this study we investigated six available techniques: MV Cone-Beam CT (S), kV Cone-Beam CT (E,V), kV CT-on-rail (S), orthogonal-angled planar imaging of kV-kV pair (E, V), MV-MV pair (E, V and S) and kV-MV mixed pair (E, V). In the case of Artiste linear accelerator, we performed a partial MV Cone Beam CT using 6-MV photon beam, with 200° arc rotation clockwise from 260° to 100°; the MV-MV 2D-planar imaging using 6 MV photon beam at gantry angles 0° and 90° for 15x15 cm² field size and helical kV-CT with tube voltage of 120 kV and 3 mm slice thickness. As for Synergy linear accelerator the imaging protocol involved a complete kV-CBCT with a tube voltage of 120 kV; the MV- MV imaging at gantry angles 0° and 90° using 6-MV photon energy, 16x16 cm² field size; the kV-kV planar imaging at gantry angles 90° and 180° with 120kV tube voltage and 16x16 cm² field size. On the TrueBeam accelerator we measured the dose of a complete kV-CBCT from 180° to -180° arc rotation with 125 kV tube voltage; the MV-MV planar imaging at 0° and 90° gantry angles with 6-MV photon energy, 15x15 cm² field size; the kV-kV planar imaging at gantry angles 0° (80kV) and 90° (100kV) besides the largest field size (20x27) and orthogonal-angled kV-MV mixed modality with 80 kV (anterior) and 6-MV (lateral) energies with 15x15 cm² field sizes. We performed point dose measurement with Farmer ionization chamber (31003 PTW) connected to a Unidos electrometer (PTW). For the measurements we used the CIRS Model 74-007 IMRT Thorax Phantom (Fig.1) which is made of tissue equivalent materials and consists of three different inhomogeneities (normal tissue, bone and lung). The ionization chamber can be placed
in different inserts in ten different points of the phantom, making possible to collect data from the ipsilateral lung, contralateral lung, heart, spine and mediastinum.
The measurements were undertaken with a phantom positioned for left sided breast radiotherapy. The isocentre was positioned within the treated breast, the same position that would be used for treatment. If the imaging could not be performed in the shifted position, we centred the couch.

**Results**
The mean dose values of all points measured in the phantom can be found in Table 1 with respect to each vendors. The only comparable result was yielded by the MV-MV imaging technique. The differences between the measured values were less than 1 cGy under the same conditions. The kV-kV 2D planar imaging was found to deliver the lowest dose (0,03cGy) and MV-CBCT imaging caused the highest dose (7,11cGy) to the whole phantom.

During the left sided breast irradiation we should take into account the dose of the heart (P4) and the ipsilateral lung (P8) during treatment planning. Fig.2 and Fig. 3 show the absorbed doses in these points. The highest values were found in the case of MV-CBCT imaging (7,4cGy and 7,3 cGy). As for kV-kV imaging, the comparison of the data measured in these two points are collected in Table 2 with respect to each machine. The absorbed dose in the spine was low (maximum 0,9cGy) in all 2-D planar imaging, as it is out of the field, and in all types of kV-CT. In case of MV-CBCT 6,1cGy was measured.

**Discussion**
One possible approach to reduce absorbed dose is to use lower energies for imaging when it is available on the system. According to our study, orthogonal-angled kV-kV paired 2-D planar imaging results in the lowest dose during verification and we can reach satisfying soft tissue contrast. The main advantage of CBCT is getting volumetric information of the patient. On the other hand CT imaging takes more time – in some cases this may be considered, according to the patient’s condition and the possible setup uncertainties. One option to replace CBCT is to use fiducial markers placed at the edge of lumpectomy cavity during surgery. These markers can also be tracked with optical imaging and planar imaging. The quality of the images were not taken into consideration in this study, but all of them are used in daily clinical routine. Before choosing the optimal verification protocol, we should take into account all of these factors.
Figure 1. The CIRS Model 74-007 IMRT Thorax Phantom and the measuring points

<table>
<thead>
<tr>
<th>Location</th>
<th>TrueBeam [cGy]</th>
<th>Synergy [cGy]</th>
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<tbody>
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<td>4. Heart</td>
<td>0.05</td>
<td>0.06</td>
</tr>
<tr>
<td>8. Ipsilateral lung</td>
<td>0.06</td>
<td>0.07</td>
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</table>

Table II. The comparison of the data measured during the kV-kV imaging
Figure 2. Absorbed doses in point 4 measured in CIRS IMRT thorax phantom with different image guidance techniques

Figure 3. Absorbed doses in point 8 measured in CIRS IMRT thorax phantom with different image guidance techniques
References:


DOSE PROFILES MEASUREMENTS FOR LEKSELL GAMMA-KNIFE MODEL C USING AUTOMATIC POSITIONING SYSTEM WITH STEREOTACTIC DIODE DETECTOR

Hrvoje Hršak¹, Marija Majer², Zdravko Heinrich¹

¹University Hospital Centre Zagreb, Kišpatićeva 12, 10000 Zagreb, Croatia
²Ruđer Bošković Institute, Bijenička 54, 10000 Zagreb, Croatia

E-mail of corresponding author: hhrsak@kbc-zagreb.hr

Introduction

Gamma-Knife radiosurgery is highly precise method for treatment of small intracranial lesions with high single radiation dose [1]. For Leksell Gamma-Knife unit (LGK) Model C (Elekta AB, Stockholm, Sweden) radiosurgery uses 201 convergent narrow Co-60 photon beams, collimated with the 18, 14, 8 and 4 mm helmet collimator.

The quality assurance program (QA) for Gamma-Knife includes periodical checks of size and position of single collimator helmet dose distributions from which complex clinical distributions are composed [2]. This is done by measuring dose distributions profiles in a three orthogonal planes (usually along x, y and z - axis of Gamma-Knife unit) and comparing them against the profiles measured during commissioning of Gamma-Knife or calculated by the treatment planning system (TPS). Various types of detectors are used for Gamma-Knife dose profiles measurements: radiographic and radiochromic films [2], small ionization chambers, diodes and diamond detectors [3,4]. In investigations published so far, for Gamma-Knife dose profiles measurements with point detectors the phantom with detector was moved manually from previous to next measuring point using trunnions [5] with resolution usually not better than 1 mm. In that measuring set-up dose profile scanning is limited to transversal direction of Gamma-Knife unit and time consuming due to manual adjustment of phantom to every measuring point. In addition, dosimetry of small beams using point detectors is associated with dosimetrical problems such as volumetric averaging of a measured signal, loss of lateral electronic equilibrium, and change in a dose response due to detector density perturbation of small photon beam.
As radiochromic film emerged as a reliable and practical dosimetry instrument with high spatial resolution, near tissue equivalence and weak energy dependence [6], idea of Gamma-Knife profiles scanning with point detectors was abandoned and replaced by film dosimetry. However, small volume detectors such as semiconductor diode are often used for dosimetry of small radiosurgery beams, because of a small measuring volume (thickness of a several tens of 1 μm and diameter equal or less than 1 mm) and therefore high spatial resolution.

In this paper we propose method for high resolution dose profiles measurements for LGK Model C using the Automatic positioning system (APS) - a motorized system for positioning of the patient head and stereotactic frame into the target coordinates [7], with p-type stereotactic diode detector. Measurements were performed in a non-reference geometry where spherical phantom with detector was moved from one measuring point to another using APS. The method was validated by comparison of profiles simulated in a non-reference geometry of phantom against the profiles simulated in a reference geometry using the Leksell Gamma Plan (LGP) TPS and by comparison of profiles measured with EBT3 film with phantom in reference geometry against the profiles measured with the Diode SRS in a non-reference geometry.

Materials and methods

In the Leksell Gamma-Knife unit Model C 201 Co-60 sources are placed in five parallel rings (marked as A, B, C, D and E ring in Fig.1) on a semi-hemispherical surface, delivering the 201 photon beams that are focused to single point of intersection, known as the unit center point (UCP), at the source-to-focus distance of 40.3 cm. The beams are collimated with a 18, 14, 8 and 4 mm helmet collimator giving almost spherical dose distribution around UCP.

All measurements were performed for 18, 14, 8 and 4 mm helmet collimator using spherical phantom made of tissue equivalent polystyrene material (Elekta AB, Stockholm), having a diameter of 16 cm. This phantom is composed of two hemi-spheres with a special inserts that can accommodate radiochromic film, ionization chamber or diode detector. The phantom was mounted to the Leksell stereotactic frame and adjusted to the APS (Fig. 2). Diode detector was positioned in the centre of the phantom, with the longitudinal axis along z – direction of Gamma-Knife, in a way that the centre of measuring volume of detector coincide with the centre of the phantom.

Prior to dose profiles measurements, phantom with stereotactic frame was scanned using multi-slice Siemens Somatom Sensation Open CT (Siemens AG Medical Solutions, Erlangen). One hundred sixty non overlapping images of phantom were acquired in stereotactic conditions. For imaging standard 120 kVp head CT protocol was used with a slice thickness of 1.2 mm. CT-images were imported into treatment planning system LGP 10.1.1 and defined in stereotactic space. To generate automatic measurement sequences for dose profiles scanning multi shot
treatment plans were generated separately for each collimator helmet (18, 14, 8 and 4 mm). Each shot represents one measuring point. For every dose profile measurements shots were distributed along principal axes of the Gamma-Knife (x or z – direction) in steps of 0.2 mm, i.e. resolution of measurements was 0.2 mm. After the treatment plan was finished treatment protocol was exported to the console of Gamma-Knife unit and measurements were performed with diode detector in a non-reference geometry of the phantom where the centre of phantom and UCP does not coincide, at the points defined by protocol (Fig 3). Measurements with radiochromic film were performed in a reference geometry of the phantom (centre of the non-moving spherical phantom coincides with the UCP).

A p-type silicon stereotactic diode (Diode SRS PTW60018, PTW-Freiburg) was used for the point measurements in this work in conjunction with PTW UNIDOS E electrometer (PTW-Freiburg). Measurement time for every point was t = 10 s. The active volume of the Diode SRS detector is a disk-shaped silicon chip with a diameter of 1.1 mm and thickness of 250 µm having nominal response of 175 nC/Gy. For the 2D measurements EBT3 radiohromic film (ISP-Wayne) was used in conjunction with the EPSON EXPRESSION XL10000 scanner (SEIKO EPSON-Nagano). EBT3 films were calibrated using 18 mm collimator Co-60 photon field following adapted calibration protocol proposed by Devic et al [6]. Film scanning resolution was 96 DPI.

The method for dose profile measurement using point detector and the APS was validated by comparison of profiles simulated in a non-reference geometry of phantom (scoring voxel size of 1 mm) against the profiles simulated in a reference geometry using the LGP TPS, and then by comparison of the profiles measured with the Diode SRS in a non-reference geometry against profiles measured with EBT radiochromic film with phantom in reference geometry.

**Results and discussion**

The Gamma-Knife dose profiles were simulated and compared for non-reference and a reference geometry using the LGP TPS for the 18 mm helmet collimator, for which the largest difference between profiles in ref. and non-ref. geometry was expected (Fig 4). Small difference between simulated profiles in ref. and non-ref. geometry was observed on the plateau of z-profile (max. 1.5 %), while x-profiles simulated in non-ref. geometry showed excellent agreement with the x-profiles simulated in ref. geometry. This indicated that the non-reference geometry in which the phantom is travelling along x or z-axis, instead of being in the centre of the Gamma-Knife unit, is a suitable for accurate dose profile scanning of Gamma-Knife profiles with point detector. Dosimetry of small fields with conventional point detectors such as ionization chamber is usually erroneous because of the volume averaging of measuring signal and density perturbation of a beam conditions. However, if the volume of detector is small enough, like in the case of diode or diamond detector, volume averaging effect in a small photon fields, such as of Gamma-Knife,
could be negligible. In this work, dose profiles for the largest, 18 mm collimator and the smallest, 4 mm collimator were measured with the resolution of 0.2 mm, using the Diode SRS detector and EBT3 film, and compared (Fig 5 and Fig 6). Because of superior spatial resolution, no volume averaging effect and tissue equivalence, radiochromic film represents common instrument for dosimetry of a small fields [6] and in this work it was used as a reference detector for checking the suitability of diode detector for Gamma-Knife profiles measurement. Excellent agreement of EBT3 and Diode SRS profiles was found for the 18 and 4 mm collimator and a small volume averaging effect for the EBT3 z-profile. This is in contrast to expected since the EBT3 is considered as a volume averaging free instrument. The finite focal spot size of the XL10000 scanner may be considered as a source of volume averaging in this case. Also, better signal to noise ratio (SNR) was observed for the Diode SRS detector, comparing to EBT 3 film.

Conclusions

Diode SRS PTW60018 detector represent good choice for the Gamma-Knife dosimetry because of high spatial resolution and good signal response and in conjunction with the Automatic positioning system it can provide high resolution and volume averaging free dose profile measurement for the Leksell Gamma-Knife Model C. In addition, less noise in a measured signal and simpler workflow represent one advantage over the radiochromic film. However, EBT3 radiochromic film remains the detector of choice for the 2D dosimetry because of high spatial resolution and tissue equivalence.

References


**Fig. 1** Helmet collimator with five rings containing beam channels which focus 201 photon beams to a single point of intersection, known as the unit centre point (UCP).
Fig. 2 The spherical phantom mounted to the 4 mm helmet collimator and adjusted to the UCP.

Fig. 3 Dose profile were measured with the Diode SRS detector in a non-reference geometry by moving the phantom with the APS along principal axes of the Gamma-Knife (x or z – direction) in steps of 0.2 mm.
Fig. 4 Gamma-Knife profiles simulated using LGP TPS for the 18 mm helmet collimator (scoring voxel size of 1 mm), for ref. and non-ref. geometry of the phantom. Small difference was observed on the plateau of z-profile, while x-profiles shows excellent agreement.

Fig. 5 Gamma-Knife $x$ and $z$-profiles for the 18 mm collimator, measured with the EBT3 radiochromic film and Diode SRS detector. EBT3 and Diode SRS profiles are in excellent agreement and no volume averaging effect is observed.
Fig. 5 Gamma-Knife x and z-profiles for the 4 mm collimator, measured with the EBT3 radiochromic film and Diode SRS detector. EBT3 and Diode SRS profiles are in good agreement and small volume averaging effect is observed for the EBT3 z-profile.
Significance beam commissioning, beam modeling and verification TPS on example
Department of Radiotherapy in Nis

Jelena Stankovic¹, Eduard Gershkevitsh², Tamara Jovanovic¹, Dragan Nikolic¹, Milos Jonic¹

1. Clinical Center Nis, Clinic of oncology, Department of radiotherapy, Serbia
2. North Estonian Medical center, Tallin, Estonia

jelenavms@gmail.rs

Introduction

Two new accelerators Elekta Synergy Platform were installed at the Department of Radiotherapy of the Clinical Center Nis in 2016. During the acceptance test two Elekta accelerators were beam-matched, according to the manufacturer recommendations. The request was to introduce these two new machines into clinical practice as soon as possible and to skip the full beam data acquisition, modelling and commissioning of the systems. The models of machines from the other Center were copied and introduced into the existing TPS XiO V4.50.00 Elekta CMS Software.

However, following the international recommendations the full beam data acquisition has been performed and models fine-tuned before being used for planning of external beam radiotherapy. Here we present our findings when comparing the beam data in the models based on other linac data with the actual measurements on the linac.[1]

Materials and methods

In order to verify TPS it was necessary to perform the beam data collection for all energies (two photon and four electrons) in both accelerators (Elekta 1 and Elekta 2). IBA 3D Blue Phantom with CCU electrometer and CC13 ionization chambers have been used for the beam data collection. IBA software OmniPro Accept 6.6 was used for recording of the characteristics of photon and electron beams. 1D Phantom (WP1D) and Dose 1 electrometer with FC56G and NACP-02 plane-parallel ionization chambers were used for non-scanned beam data acquisition from photon and electron beams, respectively.

The data were collected in accordance with the TPS manufacturer requirements.

As for the scan-data, PDDs for both photon energies 6 MV and 10 MV for the field sizes from the smallest to the largest, PDDs for all electron energies (6, 8, 10, 12, MeV) and all available applicators, open field profiles, diagonal profiles, wedged field profiles, wedged field PDDs, MLC profiles were made. For the non-scan data: absolute dosimetry, output factors, wedge factors both collimator angles, applicator factors were made.

Results
The recorded data from Elekta 1 and Elekta 2 accelerators were compared to TPS and with each other. The results are presented in tables and graphs. The data comparison was made in water for which international organisations (IAEA, ESTRO, AAPM) recommend an accuracy of 2% for the open and 3% for the wedged field in the useful (>80% isodose) part of the beam.

The following was concluded:
- For output factors: differences are smaller than 1% for field sizes larger than 4x4 cm$^2$. The differences for smaller fields could be the consequence of using different detectors for the current measurements and beam data acquisition for modeling (small field output factors were collected in the initial model with Farmer chamber input beam data problem).
- For wedge factors: the differences for average wedge factors are within 1.1%.
- For open field profiles the max difference in diagonal profiles for 10X beam is reaching 3%, indicating possible differences in flattening filters between modelled and actual machine. This however, does not affect significantly aligned profiles for 30x30 cm$^2$ field sizes and smaller. The agreement for 6X beam is good.
- Wedge field PDDs: the agreement for both energies is good. The range of differences is within [0.5%;1.1%] for 6 MV and [-0.5%;1.4%] for 10MV.
- The wedge profile of Elekta 1 accelerator has the maximum difference at d$_{max}$ of 4.4% for both energies. This difference is reduced to 2.5% and 1.9% at depth of 10 cm for 6 and 10 MV beam, respectively, indicating the differences in the modelled wedge profile and actual measurements. To compare this in absolute terms the wedge profiles would also need to be scaled by the wedge factor (+1.1% difference). Elekta 2 wedge profiles have better agreement.
- MLC profiles: the asymmetrical (2.5 and 7.5 cm across center axis) MLC only field were created and profile scans were performed and compared with calculations. As showed in the Figure 1, the modeling parameters for 6X beam for MLC and back-up jaw were swapped which lead to incorrect profile representation in the TPS (beam data input problem).

Discussion

All the data collected were sent for analysis to a consultant physicist (Gershkevitsh E.) with experience in the field of QA programme and TPS verification procedures. As seen in the conclusion, most of the analyzed data are within the above-mentioned tolerance. However, we decided to insert the measured beam data into TPS and redo the modeling to further minimize the discrepancy between calculated and measured doses. So, during the consultant’s visit the following was carried out:
1. Two universal models for photon beams (6X and 10X) for both machines were created instead of four “individual”
2. Absolute output values in the model were adjusted to reflect the actual measured PDD value at 10 cm depth (100 cm SSD and 10x10 cm$^2$) value was based on the other unit beam quality
3. The output and wedge factors from both units (Elekta 1 and Elekta 2) were averaged and average value inserted into TPS.
4. PDD curves for open and wedged beams were transferred to the TPS and spectrum of the model refitted.
5. MLC and back-up jaw transmission values changed according to the measurement results.
6. Wedge profiles were transferred and wedge profile shape refitted according to the measured profiles.

When we finished re-modeling of TPS, the models were tested by using IAEA TECDOC 1583 methodology and CIRS phantom on both units. The test case description and results are shown below in the Table 1. [2]

From the presented test results in TPS, it can be seen that all the cases passed the test except the case with the wedge.

Unfortunately, we were not able to do the TPS testing before the re-modeling, but if you take into consideration the shown deviations we can assume that in some cases we would have summed them up and would be outside the tolerance, so anyway we would not have had such a good test results.

In the process of implementation of the external beam therapy, each step, starting from beam scanning to the treatment could bring in an error. The QA procedures, are aimed to minimize those mistakes as much as possible. One of the steps in this process is TPS modelling. Commissioning is important part of QA program and it should be a good practice to use TPS models with the measured data from the actual unit and TPS should be verified before the treatment start.

References

### Table 1

<table>
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<tr>
<th>Case</th>
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<th>Meas. point</th>
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<th>Measurement results (Gy)</th>
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<td>-0.2</td>
<td>3</td>
<td>P</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>F3</td>
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<td></td>
<td></td>
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</tr>
<tr>
<td></td>
<td>SUM</td>
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<td>3</td>
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<td>0.028</td>
<td>-1.7</td>
<td>4</td>
<td>P</td>
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<tr>
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<td></td>
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<td></td>
<td></td>
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<td>0.032</td>
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<td></td>
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<td>0.385</td>
<td>-2.0</td>
<td>3</td>
<td>P</td>
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</table>

**Figure 1**
| 5 | Customised blocking | | 6 | L-shaped field | | 7 | Plan with assymetric fields and wedges | | 8 | Plan with non-coplanar fields |
|---|-------------------|---|---|-------------------|---|---|---|
|  | F2 | 0.019 | 0.029 | -2.1 | 4 | P | |  | F1 | 1.000 | 0.998 | 0.2 | 2 | P | |  | F1 | 1.000 | 1.008 | -0.8 | 3 | P |
|  | F3 | 0.675 | 0.691 | -3.2 | 3 | FAIL | |  | F2 | 0.500 | 0.512 | -2.3 | 4 | P | |  | F2 | 0.500 | 0.496 | 0.8 | 3 | P |
|  | F4 | 0.019 | 0.029 | -2.0 | 4 | P | |  | F3 | 0.500 | 0.501 | -0.2 | 4 | P | |  | F3 | 0.500 | 0.499 | 0.2 | 3 | P |
| | SUM | 1.088 | 1.134 | -2.3 | 3 | P | | | SUM | 2.000 | 2.011 | -0.5 | 3 | P | | | SUM | 2.000 | 2.003 | -0.1 | 3 | P |
|  | 2 | 2.000 | 1.975 | 1.2 | 3 | P | |  | 3 | 2.000 | 1.974 | 1.3 | 3 | P | |  | 5 | 1.711 | 1.698 | 0.6 | 5 | P | |  | 5 | 1.711 | 1.698 | 0.6 | 5 | P | |  | 5 | 1.711 | 1.698 | 0.6 | 5 | P |
|  | 7 | 1.670 | 1.655 | 0.7 | 4 | P | |  | 7 | 1.258 | 1.237 | 1.1 | 5 | P | |  | 10 | 1.711 | 1.698 | 0.6 | 5 | P | |  | 10 | 1.711 | 1.698 | 0.6 | 5 | P | |  | 10 | 1.711 | 1.698 | 0.6 | 5 | P | |  | 10 | 1.711 | 1.698 | 0.6 | 5 | P |
Experimental determination of the standard IAEA holder correction factor for RPLD postal dose audit applications.

Pavel Kazantsev, Paulina Wesolowska, Tania Santos, Domonkos Szegedi, Joanna Izewska

*International Atomic Energy Agency, Dosimetry and Medical Radiation Physics Section, Vienna, Austria*

Corresponding author e-mail address: p.kazantsev@iaea.org

**Introduction**

In 1969-2016, radiotherapy level dosimetry audit service provided by the International Atomic Energy Agency (IAEA) and the World Health Organization (WHO) utilized thermoluminescence dosimeters (TLDs) which were positioned in a water phantom using the standard IAEA holder for irradiation in the reference conditions [1]. From 2017, after comprehensive commissioning process, the IAEA Dosimetry Laboratory introduced a radio-photoluminescence dosimetry (RPLD) system into the full-scale operation. This article describes the methodology and the results of the RPLD holder correction factor determination.

**Materials and methods**

The holder correction factor is required to account for the change of the dosimeter signal due to radiation beam attenuation and scatter in the holder walls. It is defined as the ratio of the absorbed doses to water evaluated from the readings of dosimeters positioned inside a water phantom without and with the holder in the same irradiation conditions. The IAEA standard holder is used for RPLD irradiations. It is a 26 cm long PMMA tube with 10 mm outer diameter and 2 mm wall thickness which has an opening for a dosimeter at 10 cm distance from the tip; 3 legs and an additional metal ring allow precise and stable positioning of the holder inside a water phantom. Some audit participants also utilize older version of the holder with the opening at 5 cm from the tip and a 5 cm long distance plug which makes available both 5 and 10 cm depth setups corresponding to the reference conditions described in TRS-398 for different beam qualities [2]. For the 10 cm setup, the distance plug is removed after the holder positioning. It was shown [3, 4] that the corresponding holder correction factor is essentially the same for both 5 cm and 10 cm setups. Therefore, correction factors are needed for two types of holders, in the range of photon beam energies used for radiotherapy treatments.

The magnitude of the holder correction, among other things, depends on the part of the dosimeter sensitive volume covered by the holder walls and is different for TLDs and RPLDs. The standard IAEA TLD capsule is 23 mm long and it has a 19 mm long inner compartment filled with TLD powder. Therefore, a substantial part of the TLD’s sensitive volume is shielded by the holder walls. The RPL dosimeter is a 12 mm long glass rod placed in a plastic capsule with the outer dimensions similar to the TLD capsule, however, the signal is taken only from the 6 mm long central area of the rod. It coincides with the holder inner diameter and minimizes the holder shielding effect but also makes the RPLD capsule susceptible to potentially increased measurement uncertainty in case of inaccurate positioning inside the holder tube (Figure 1 a, b).
The holder correction factor can be determined by direct comparison of the absorbed doses derived from the signal of RPLDs irradiated using the holder and without it; otherwise RPLD can be replaced by a small ionization chamber with a similar sensitive volume [3]. Also the holder correction factor can be calculated using Monte Carlo method [4].

A new method developed for the holder correction determination involves the measurement of two beam profiles (profile pair) in a water phantom: 1) with a small volume dosimeter passing through the opening of the holder (Figure 1 c) and 2) the profile in the holder absence at the same depth. The integrals of the area under the beam profiles within the central 6 mm distance represent the dose averaged within the sensitive volume of RPLD for both scenarios. The ratio of integrals with and without the holder gives the holder correction factor.

The dosimeters tested in this study included a PTW 60019 microDiamond, IBA Razor diode and PTW 31014 PinPoint. Some of their characteristics relevant to the current study are presented in Table 1. Initial testing of the dosimeters’ suitability was performed in a 10×10 cm² field of a Co-60 beam at SSD=100 cm and 10 cm depth. It included the measurement of a profile pair per dosimeter and aimed to evaluate the dosimeter with the least averaging effect and noise.

The dosimeter considered the most suitable was subsequently used for the measurement of profile pairs in two scanning directions and two dosimeter orientations. The holder correction calculation was based on the profile pair obtained through averaging of all measured profile pairs.

Following the initial study using a Co-60 beam, the RPLD holder correction factors were also obtained for 6 MV, 10 MV and 18 MV beams. The method was benchmarked against the determination of the well-known TLD holder corrections [3, 4] using relevant to TLDs 19 mm integration limits within the measured profile pairs.

As it was previously noted, the misalignment of the RPL dosimeter readout area with the holder tube may potentially lead to an increased uncertainty in the absorbed dose to water evaluation. The magnitude of the effect was estimated by calculating the holder correction factors from the measured profile pairs with the integration limits shifted from the axis position. The overall uncertainty, related to both the holder correction factor determination methodology and the RPLD positioning in the holder, was evaluated.

Results

Results of the initial dosimeter testing are presented in Figure 2. Following this study, the microDiamond dosimeter was chosen for the subsequent measurements.

The holder correction factors are presented in Figure 3. Data for TLDs is compared to the previously determined holder corrections with the uncertainty of 0.3% [3]. The RPLD holder correction factors dependence on the beam quality can be approximated with linear equations:

- \( y = -0.0006 \times D_{20}/D_{10} + 1.0035 \) for the holder with opening at 5 cm depth
- \( y = -0.0044 \times D_{20}/D_{10} + 1.0087 \) for the holder with opening at 10 cm depth
Results of the holder correction factor determination for the different RPLD shifts during irradiation are presented in Figure 4.

The uncertainty of the RPLD holder correction was estimated as type A with the magnitude of 0.08%. At the same time, the dosimeter positioning sensitivity gives an additional input in the uncertainty budget. Assuming 0.5 mm potential displacement, a maximum 0.40% increase of the holder correction factor is expected for a Co-60 beam. Correspondingly, applying the rectangular distribution of the positioning probability, the resulting additional uncertainty of the holder correction factor is 0.12%. Therefore, the overall uncertainty is 0.14%.

Discussion

The initial dosimeter testing showed that a microDiamond and a Razor diode expressed the lowest volume averaging effect but the signal of the latter one contained more noise. The profile measured with a PinPoint ionization chamber was very smooth but, due to comparatively large dosimeter volume, the associated signal averaging was significant.

The comparison of the newly calculated TLD holder correction factors with the existing ones showed good agreement within 0.15% with each other which proved the applicability of the methodology.

The calculated RPLD holder correction factors did not exhibit significant beam quality dependence. For the standard IAEA holder, the correction factor ranged from 1.007 to 1.006 for Co-60 and 18 MV beams respectively. For the holder with the opening at 5 cm, it had the same value of 1.003 for Co-60 and 6 MV beams.

The intrinsic uncertainty of the measurement method was lower than the one associated with the dosimeter positioning in the holder. One should distinguish it from the setup error when a greater shift of the capsule may occur due to a user mistake. Displacement of the dosimeter by 2 mm during the irradiation at 10 cm depth, even if the holder was set up correctly, may lead up to 1.7% error in the absorbed dose to water evaluation for a Co-60 beam. It clearly shows that precise positioning of the dosimeter inside the holder opening is crucial for the accurate interpretation of the audit results.

References


Figure 1. Correct setup with the RPLD readout area aligned with the holder tube (a) and incorrect setup with the RPLD shifted by 1 mm (b). Through-holder beam profile measurement (c).
Figure 2. Initial dosimeter testing in a Co-60 beam: profile pairs obtained by microdiamond (a), Razor diode (b) and PinPoint 31014 (c).

Figure 3. Holder correction factors for the holders with the opening at 5 and 10 cm, calculated for TLD (a) in comparison with the existing corrections [3, 4] and RPLD (b).
Figure 4. Change in the RPLD holder correction factors due to the dosimeter positioning shift for the IAEA holder with the opening at 10 cm depth.

Table 1. Characteristics of the dosimeters tested.

<table>
<thead>
<tr>
<th>Dosimeter</th>
<th>Outer dimensions</th>
<th>Sensitive volume dimensions</th>
<th>Nominal response</th>
<th>Radiation incidence direction</th>
<th>Typical use</th>
<th>This study</th>
</tr>
</thead>
<tbody>
<tr>
<td>PTW 60019</td>
<td>Diameter 7 mm, length 45.5 mm</td>
<td>Diameter 2.2 mm, thickness 1 μm</td>
<td>1 nC/Gy</td>
<td>Axial</td>
<td>Radial</td>
<td></td>
</tr>
<tr>
<td>microDiamond</td>
<td>Diameter 4 mm, length 60 mm</td>
<td>Diameter 0.6 mm, thickness 20 μm</td>
<td>4.1 nC/Gy</td>
<td>Axial</td>
<td>Radial</td>
<td></td>
</tr>
<tr>
<td>IBA Razor diode</td>
<td>Diameter 3.4 mm, length 57 mm</td>
<td>Diameter 2 mm, length 5 mm</td>
<td>0.4 nC/Gy</td>
<td>Axial and radial</td>
<td>Radial</td>
<td></td>
</tr>
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</table>
Setup position errors in image-guided head-and-neck radiotherapy with respect to the reference isocentre position

Helena Lenko, Primož Peterlin. Institute of oncology Ljubljana, Zaloška c. 2, SI-1000 Ljubljana, Slovenia. ppeterlin@onko-i.si

Introduction

Setting the reference isocentre (sometimes also called the patient origin) is one of the key steps in the CT simulation process. Reference isocentre is a point fixed to the patient body, relative to which the treatment field isocentre (when isocentric technique is used) is defined. The reference isocentre is neither the DICOM origin, which is defined with respect to the CT scanner, nor the treatment field isocentre, which is usually not yet known at the time of CT simulation. A well-chosen reference isocentre fulfills several criteria: (a) it is placed over stable anatomical structures, and (b) it lies as close as possible to the presumed treatment field isocentre. In head-and-neck radiotherapy, reference isocentre is set either to the patient skull or to the patient neck. The first choice has the advantage of being placed over a very stable anatomical structure, but it lies further away from the treatment field isocentre. Placing the reference isocentre onto the patient neck has the advantage of being closer to the treatment field isocentre, but this point may not be accessible, e.g., when tracheotomy was performed. The purpose of this study is to determine whether either choice of reference isocentre is advantageous to the other with respect to patient set-up errors, and consequently establish the priority of reference positioning criteria.

Materials and methods

Set-up correction data for 117 patients with head-and-neck cancer treated with volumetric modulated arc therapy (VMAT) in the years 2013 and 2014 on a single treatment machine (Varian NovalisTx; Varian Medical Systems, Palo Alto, CA, USA) in our clinic were analysed. All patients were immobilised with a thermoplastic mask (Posicast 5-point head, neck and shoulder mask; Civco Medical Solutions, Orange City, IA, USA), radio-opaque markers were taped to the thermoplastic shell to designate reference isocentre, and a planning CT scan was taken on a CT scanner with a flat-top couch.

At every treatment fraction, a patient was positioned on the treatment couch using the markers as guidance so the patient reference isocentre was aligned with the in-room lasers. Then, the planned shifts in the longitudinal, lateral and vertical direction were performed, bringing the treatment field isocentre in alignment with the in-room lasers. Two orthogonal kV images were then taken using the on-board kV imager (On-Board Imager (OBI); Varian Medical Systems). Focusing on bone anatomy landmarks, the radiation therapist in charge compared the alignment with the digitally reconstructed radiographs (DRR), determined the necessary corrections in patient position and refined patient position. All the data concerning the positional corrections...
applied at every treatment fraction are recorded in the ARIA Oncology Information System (Varian Medical Systems). Using the Statistics/Trends tool in the Offline Review application, the positional corrections (Online OBI Match Results) were exported to a text file. Further processing was performed using in-house scripts written in GNU R. From the collected data, group systematic error, its standard deviation, and random error were calculated (Remeijer et al., 2000).

Results

Altogether, 3423 records on positional corrections for 126 treatment plans created for 117 patients were analysed. 952 records contained gross errors – positional displacement larger than 5 mm. 10 records contained positional displacement larger than 15 mm and were excluded from the further analysis. In 76 of the patients, reference isocentre was set on the patient skull, and in 41 of them, on the patient neck. As some of the patients were treated with more than one treatment plan, the total numbers of treatment plans with the reference isocentre on the patient skull and neck were 79 and 47, respectively. Fig. 1 shows the distribution of longitudinal shifts for both groups of treatment plans.

Table 1 shows the systematic error $M$, its standard deviation $\Sigma$, the random error $\sigma$, and the total error $s$ of patient position in vertical (Vrt), longitudinal (Lng), lateral (Lat) direction and of the couch rotation (Rtn), with respect to the position of reference isocentre (skull/neck). Positive values refer to shifts in the inferior, cranial and left direction. In addition to the data obtained from the record & verify system, the magnitude of the 3D positional correction ($R$) was calculated, $R = \sqrt{x^2 + y^2 + z^2}$, where $x$, $y$ and $z$ are positional corrections in the longitudinal, lateral and vertical directions. Except for the systematic error in couch rotation, where the choice of reference isocentre position affects the sign of $M$, it can be seen that the magnitude of error depends more strongly on the direction of the shift than on the position of the reference isocentre, e.g., for both reference isocentre positions, the largest systematic error $M$ is in vertical direction.

In each of the four degrees of freedom (three translations plus rotation), the distributions of positional shifts for the two groups of patients – those with the reference isocentre on the skull and those with the reference isocentre on the neck – were compared using the nonparametric Wilcoxon rank-sum test (Mann-Whitney $U$-test). Wilcoxon rank-sum test shows that the distributions of positional corrections in the longitudinal direction with the reference isocentre on the skull and on the neck differ significantly ($p < 0.05$, two-tailed). In the other three comparisons of positional correction distributions, Wilcoxon rank-sum test does not show any significant difference.

The correlations between positional corrections in different directions were also examined.
Overall, positional corrections in different directions were found to be only weakly correlated, the highest being the correlation between couch rotation and the lateral translational shift for both reference isocentre set-ups, and between longitudinal and vertical shift for the patients with the reference isocentre on patient head only. Very weak correlation with the rotational correction was largely influenced by the fact that with both reference isocentre set-ups, no rotational correction was performed in approximately 75% of all cases. If only the treatment fractions in which couch rotations were performed are considered – this means excluding all treatment fractions with couch rotation in the $[-0.1^\circ, 0.1^\circ]$ range – we find out that some correlation coefficients undergo a sizable increase. The correlation coefficients between couch rotation and lateral shift increase to 0.37 ($p < 0.01$) and 0.33 ($p < 0.01$) for the reference isocentre on the skull and on the neck, respectively.

**Discussion**

Table 2 presents the results of set-up error assessment from several recent studies, employing a variety of imaging techniques. The standard deviation of the systematic error $\Sigma$ for translational errors obtained in this study is in line with the higher values published, and the values for random error $\sigma$ fit in the middle of those published. Less data has been published on rotational errors, but again, our values seem to be very close to the published ones.

Comparing to other studies, we observe a disproporionately large amount (28%) of gross errors ($> 5$ mm) in our study. We suspect the reason for this is high workload ($< 15$ min per patient fraction) coupled with the radiation therapists' confidence in the online imaging protocol, which allows them to correct the patient position before treatment.

Set-up errors are important information for deriving the CTV-PTV margin. The values for set-up errors obtained in this study are directly applicable should we decide to abolish daily imaging in favour of some average patient shift-based scheme. If we however want to stay with the daily imaging, it is the intra-fraction shifts, not the inter-fraction, which are more relevant for safety margins. Assessing the intra-fraction shifts involves additional patient imaging, which is a mildly invasive procedure. At present, we wanted to keep this study completely non-detrimental for the patient.

We have demonstrated that bringing the reference isocentre and the treatment field isocentre closer together, longitudinally, changes the distribution of position shifts in the longitudinal direction to the degree that the Wilcoxon rank-sum test shows a statistically significant difference; the difference is nevertheless small and not relevant for clinical use. All in all we can conclude that based on the results presented, neither reference isocentre set-up has shown a clear advantage over the other in terms of interfraction set-up error.
References


Table 1: Systematic and random errors in patient position with respect to the position of reference isocentre. $M, \Sigma, \sigma$, and $s$ denote the systematic error, its standard deviation, random error, and total error, respectively. Vrt, Lng, Lat and Rtn denote the vertical, longitudinal, and lateral direction, and couch rotation. All translational positional shifts are expressed in cm, and the rotation in degrees.

<table>
<thead>
<tr>
<th></th>
<th>$M_{\text{skull}}$</th>
<th>$M_{\text{neck}}$</th>
<th>$\Sigma_{\text{skull}}$</th>
<th>$\Sigma_{\text{neck}}$</th>
<th>$\sigma_{\text{skull}}$</th>
<th>$\Sigma_{\text{neck}}$</th>
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Table 2: Comparison of the values of systematic ($\Sigma$) and random error ($\sigma$) in the vertical, longitudinal and lateral direction, and couch rotation, in a few recent studies analysing set-up errors in treating head-and-neck cancer.

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<th>$\sigma$ (mm)</th>
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Evaluation of a clinically implemented adaptive radiation therapy strategy for cervix cancer

Martin Buschmann, Katarina Majercakova, Alina Sturdza, Richard Pötter, Dietmar Georg, Yvette Seppenwoolde

Department of Radiation Oncology and Christian Doppler Laboratory for Medical Radiation Research for Radiation Oncology, Medical University of Vienna, Austria

martin.buschmann@meduniwien.ac.at

Introduction

For locally advanced cervical cancer the current treatment standard is external beam radiation therapy (EBRT) followed by brachytherapy (BT) with concomitant chemotherapy. Three-dimensional conformal EBRT is increasingly replaced by IMRT and VMAT, but large deformations of the cervix-uterus necessitate margins up to 3 cm [1], which partly reduces the advantages of IMRT dose distributions. By using adaptive radiotherapy (ART) margins may be optimized and decreased.

Online ART is a promising technique for pelvic malignancies since large variations in daily anatomy occur due to random differences in organ filling (rectum, bladder, bowel). Considering the limitations of existing technology, a plan selection concept based on preplanned library of treatment plans, also known as plan-of-the-day (POTD), is currently considered a feasible ART approach for pelvic tumors. Based on online cone beam CT (CBCT) imaging the most appropriate treatment plan can be selected.

A motion model that correlates cervix-uterus and bladder movements based on two CT images with variable bladder filling was presented by Bondar et al. [2]. This makes treatments for subranges of the full motion through plan libraries feasible. In contrast to a high number of in-silico planning studies on cervix ART [3], data on clinical implementation has been published only once to our knowledge [4]. We present for the first time a clinically implemented POTD cervix ART workflow in supine treatment position and performed a dosimetric evaluation of this technique.

Materials and methods

Sixteen patients were included in a prospective study on cervix cancer ART from October 2015 until January 2017. The dose prescription for all patients was EBRT of 45 Gy (1.8 Gy/fx), followed by four fractions of BT (7 Gy/fx). Seven patients received simultaneous integrated boosts of pathological lymph nodes.

For treatment planning, two CT scans were acquired for each patient in supine position, whereas one scan was done with a full bladder and one scan with an empty bladder. An additional MR scan was performed without organ filling instructions. The three images were rigidly registered based on bony structures and the cervix-uterus (CTV tumor, CTV-T) was delineated on all scans. Based on the planning images with different bladder volumes, a decision was made if the patient is a ‘mover’ or a ‘non-mover’. Movers were patients for whom the tip of uterus moved more than
Patients with less movement were categorized as non-movers. For non-movers just one fixed treatment plan sufficed. For mover patients, the ART approach based on a POTD plan library was applied. Figure 1 shows a sketch of the clinically implemented workflow. The lymph node target volume (elective CTV) and the organs at risk (OAR) were delineated on the full bladder CT scan.

For patients treated with ART, a two stage plan library (based on a bladder volume-dependent cervix-uterus motion model) was created: one plan for empty up to mid-full bladder (empty plan), one plan for mid-full up to full bladder (full plan) and one additional motion-robust backup plan which includes all movement (robust plan). To construct the internal target volumes (ITV) a half full mid uterus structure was needed. It was generated by delineation of an additional volumetric diagnostic image with suitable bladder volume or by linear interpolation between the two extreme positions from the scans using a structure-based organ interpolation tool developed in MATLAB [5].

To account for additional organ deformation and setup errors, individualized anisotropic safety margins (5-15 mm) were added to the subrange ITVs. According to this procedure three PTVs were generated per patient: full-PTV, empty-PTV and robust-PTV. Figure 2 shows the margin and ART concepts for two patients. The full bladder CT was used for VMAT treatment planning of all plans in the Monaco 5 TPS (Elekta). Each plan was individually optimized to reach a target coverage of V95% >= 95% in the respective PTV. All treatments were delivered by an Elekta Synergy or Versa linac with 10 MV beams.

A CBCT scan was performed before every fraction, and the image was matched to the planning CT scan. The 95% isodose curves (=42.75 Gy) of the three plans in the library were superposed on the CBCT image and the best fitting plan was then chosen online (see Fig. 2). The motion-robust backup plan was only selected, if the other two plans were both suboptimal in terms of target coverage. The daily plan selection was recorded for each fraction and selection frequencies were calculated over all patients.

For the assessment of daily delivered doses to normal tissue and target structures, the rectum, bladder and CTV-T were delineated on each CBCT scan of all nine movers. The planning dose distribution of the daily applied plan was mapped onto the CBCT image without modification and daily DVH metrics were extracted: V42.75Gy for the CTV-T, V42.75Gy, V40Gy and V30Gy for bladder and rectum. Additionally the total irradiated volumes at 42.75 Gy, 40 Gy, 30 Gy and 20 Gy were recorded.

To compare the dose delivered with the POTD ART protocol to a non-adaptive approach, a non-ART scenario was simulated. In the non-ART simulation the motion robust plan was selected for all fractions. Daily DVH values of ART and non-ART dose distributions were compared by performing a paired, two-sided Wilcoxon signed-rank test (p<0.05).

Results

Of the 16 patients recruited in total, nine were declared movers (ART patients) and seven patients were classified as non-movers. Two hundred twenty-five (9x25) fractions delivered according to the POTD protocol were analyzed. Figure 3 presents the plan selection frequencies
over the treatment course. Notable is the increase in the number of fractions delivered by the empty plan from 49% in the first treatment week to 78% in the last week.

The analysis of delivered target doses revealed that the CTV-T was underdosed in 12 fractions, meaning that the target coverage V42.75Gy was lower than 95%. In the non-ART simulation sufficient target coverage was not reached in eight fractions and no statistical difference could be detected in V42.75Gy of CTV-T between ART and non-ART.

The daily total irradiated volumes were decreased significantly by ART compared to non-ART with a median difference of 87, 74 and 65 ccm in V42.75Gy, V40Gy and V30Gy, respectively. No difference in V20Gy was detected.

Scatter plots of the daily volumes V42.75Gy, V40Gy and V30Gy of rectum and bladder are depicted in Figure 4. The non-ART dose is given on the horizontal axis and the delivered ART dose on the vertical axis, which means data points below the unity line present fractions with superior organ sparing by adaption. All three DVH metrics were significantly different between ART and non-ART for both rectum and bladder. Improved organ sparing by ART was especially evident at high doses in the bladder.

Discussion

The clinical implementation of an online ART strategy for cervical cancer was described and evaluated in terms of normal tissue sparing. This protocol is based on a library of pre-planned VMAT plans and adaption decisions are based on online CBCT imaging. Only the patients with a large extent of target motion were selected for adaptive treatments, which constituted 56% of the total patient cohort.

Patients are routinely instructed to have a full bladder at the time of treatment since this anatomical configuration usually exhibits improved organ sparing. However, the plan selection statistics showed that bladder filling decreased over the course of treatment, which may be explained by early bladder radiation response or patient noncompliance.

The dosimetric evaluation showed that irradiated volumes could be decreased significantly by ART compared to a non-ART scenario while maintaining sufficient target coverage. Bladder sparing could be improved in the high dose region by ART in many fractions, but in general rectum and bladder sparing was similar for ART and non-ART in most fractions.

To justify the increased workload of an ART protocol, the benefit in terms of outcome and toxicity should be assessed in future studies and more automated software solutions are needed for a widespread adoption of ART.

References


Figure 1: Scheme of the implemented workflow

Figure 2: Adaption concept for a mover patient (upper row) and a non-mover patient (lower row). Left: The cervix-uterus (blue) and bladder (yellow) contours of the two CT scans are superimposed. Middle: For the mover a uterus mid position (purple) was interpolated and a full (orange) and an empty PTV (red) was generated. For the non-mover just one PTV was used.
Right: CBCT scan on which the 95% isodose of the best fitting plan is overlaid.

Figure 3: Plan selection frequencies over the treatment course averaged over all mover patients.

Figure 4: Scatter plots of daily volumes V42.75Gy, V40Gy and V30Gy of bladder and rectum. Delivered DVH values are plotted against simulated non-ART values. Each fraction is represented by a circle and the unity line is included.
The Multivariate Gaussian: a new approach to multichannel radiochromic film dosimetry

Ignasi Méndez and Aljoša Polšak

Department of Medical Physics, Institute of Oncology Ljubljana, Zaloška cesta 2,
Ljubljana 1000, Slovenia

E-mail: nmendez@onko-i.si

Introduction

Radiochromic films in combination with a flatbed scanner is the dosimetry system of choice for many applications in radiotherapy and radiology. Flatbed scanners can deliver three simultaneous measures of the dose distribution: one for each color channel (red, green and blue). Several methods have been proposed in the literature that combine all three measures into a single more accurate dose distribution. State-of-the-art methods for multichannel film dosimetry are perturbation models [1]. These models consider that differences between each channel dose distribution and the real dose distribution absorbed by the film are caused by small local perturbations of the dosimeter response. Perturbations are correlated between the channels. Different models postulate different correlations and, more generally, different properties of the perturbations.

The purpose of this study was to propose a new approach to multichannel radiochromic film dosimetry: the Multivariate Gaussian method; and to compare it against perturbation methods. The Multivariate Gaussian method considers that, for any dose, the probability density function (pdf) of the channel responses follows a multivariate Gaussian distribution.

Materials and Methods
Five Gafchromic EBT3 films (Ashland Inc., Wayne, NJ) from lot 06061401 where situated on top of the IBA (IBA Dosimetry GmbH, Germany) MatriXX detector inside the IBA MULTICube phantom. The phantom was set up at source-axis distance from a Novalis Tx accelerator (Varian, Palo Alto, CA). The beam energy was 6 MV and the field size was 20 cm x 20 cm. Films were irradiated with doses 1, 2, 4, 8 and 16 Gy, respectively. Doses were simultaneously measured with the MatriXX detector.

Films were scanned prior to and 24 h following irradiation. The scanner was warmed up before readings. A 4.5 cm x 20.3 cm strip of an unexposed film was fixed on the scanner bed and scanned beside the films to correct deviations of the scanner's repeatability. A 3 mm thick glass sheet was placed on top of the films to keep constant the distance between film and light source. Ten scans were taken each time for each film. In order to ensure the stability of the scanner lamp, the average scan was calculated with the last five images of each film, the first five were discarded. The lateral artifact was corrected using the model proposed by Lewis and Chan [2], the doses measured by the MatriXX detector, and the sensitometric curves for all three channels. The sensitometric curves were calculated using regions of interest centered on the field with dimensions 3 cm x 3 cm and cubic spline interpolation.

In this way, associated with each pixel of each film were: the measured dose, the pixel values (PVs) of each color channel before irradiation, and the PVs after irradiation, all PVs corrected by scanner's repeatability and lateral artifact.

Six algorithms were employed to calculate the dose from the PVs and compare it with the measured dose: the method proposed by Mayer et al.[3] using PVs (after irradiation) and net optical densities (NOD), a channel independent perturbation (CHIP) [1] method with normal perturbation's pdf using PVs and NOD, the Multivariate Gaussian (MG) method using PVs after irradiation, and the MG method using PVs before and after irradiation.

For a given dose, the conditional probability of the PVs according to the MG method can be expressed as:

\[
P(v \mid D) \sim N(\mu, \Sigma),
\]
where $\mu$ represents the mean vector and $\Sigma$ the covariance matrix. Following the Bayes's theorem, the conditional probability of the dose given the PVs is:

$$P(D \mid v) P(v) = P(D) P(v \mid D).$$

Considering the doses equiprobable, the dose pdf given the PVs is proportional to the PVs pdf given the dose. Mean vectors and covariance matrix can be obtained during the calibration for a set of doses, we interpolated them for the rest of doses using cubic splines.

Thus, given the PVs measured in a pixel, the MG yields the pdf of the dose. It is then straightforward to obtain the most probable dose and its uncertainty.

**Results**

Fig 1. and Fig.2 show 2D probability density functions for two different combinations of channel responses. For every combination of two channels, similar (approximately bivariate Gaussian) distributions were found.

Fig 3. Shows relative dose differences between measured and calculated dose distributions using the Multivariate Gaussian method with irradiated and non irradiated channels. To avoid uncertainties derived from steep dose gradients, only pixels with measured doses between $\pm 5\%$ of the prescribed dose were evaluated.

The dose uncertainty, i.e., the standard deviation of the dose difference between measured and calculated doses, obtained for each method were: 1.6, 1.6, 1.6, 1.4, 1.3 and 1.1 \% for the Mayer PV, CHIP PV, Mayer NOD, MG irradiated channels, CHIP NOD, and MG irradiated and non irradiated channels, respectively.

**Conclusion**
This work presents a novel method for multichannel radiochromic film dosimetry: the Multivariate Gaussian method. For the lot under study, the Multivariate Gaussian method was found to provide more accurate doses than the Mayer and the CHIP methods.

References


Figures

Fig 1. Probability density function 2D of PVs for each dose value. Irradiated red and green PVs are represented in the X and the Y axis, respectively. PVs are shown as differences between PVs and the mean PV for those channel and dose.

Fig 2. Probability density function 2D of PVs for each dose value. Irradiated red and non-irradiated blue PVs are represented in the X and the Y axis, respectively. PVs are shown as differences between PVs and the mean PV for those channel and dose.
Fig 3. Relative dose differences between measured and calculated dose distributions using the Multivariate Gaussian method with irradiated and non irradiated channels.
Implementation of daily QA programme for IGRT linear accelerators-first experiences

Milana Marjanović, Laza Rutonjski, Borislava Petrović, Ozren Čudić

Institute of oncology Vojvodina, Sremska Kamenica

E-mail address: milana.marjanovic92@gmail.com

Introduction: The use of radiation therapy techniques, such as IMRT and VMAT, which are extremely precise, enables most precise delivery of the treatment to the patient. To achieve that, patient must be fixed in treatment position during the overall treatment. Introduction of IGRT requires implementation of quality control testing of the equipment, as precision became of most crucial importance. Mechanical and geometrical precision of imaging systems directly affects positioning and delivery of the treatment.

Materials and methods: Institute of oncology Vojvodina holds two Versa HD machines, manufactured by Elekta, equipped with MV and kV imaging systems. The daily testing routine is based on the use of ISO Cube (plastic water phantom), which contains one central marker, one offset marker and external concentric circles. These are used for isocenter checks, by ISO Analyze program, that actually analyses DICOM images. ISO Cube phantom is set on a treatment couch and align offset marker to room lasers. Images of a phantom are taken by CBCT and based on the images, automatic table movement is checked. Furthermore, center of the phantom is set to room lasers and images are taken with XVI and EPID. These images are used to check collimator, gantry and couch rotation isocenter, and also MV radiation isocenter, 2DkV isocenter, kV-CBCT isocenter and mechanical isocenter.

Results and discussion: The results were presented as the difference between isocenters (collimator, gantry, couch rotation isocenter, MV radiation isocenter, 2DkV and kV-CBCT isocenter) and central marker; and as the difference between mechanical isocenter and central marker, MV radiation isocenter, 2DkV and kV-CBCT isocenter (Table 1.). After initial setup, result are tracked continuously. Currently, they are limited, but so far results of mechanical and geometrical tests are in good agreement with the results that are actually expected. They were within the tolerance of 2 mm, and in agreement with AAPM recommendations. So far, the measurement principle is simple and convenient. The results give confidence that equipment performs effectively well on daily basis.

References:


“Quality assurance for kilovoltage cone beam computed tomography”, Joerg Lehmann, Stanley Skubic

“ISO Cube daily QA phantom”, CIRS

Table 1. The results of mechanical and geometrical tests for seven consecutive measurements

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Monaco MLC geometry modelling for older models of Varian linacs

Rihard Hudej, Aljaša Jenko, Sašo Pulko, Matevž Mlekuž

Department of Radiation Oncology, Section of Medical Physics, Institute of Oncology, Ljubljana, Slovenia
rhudej@onko-i.si

Introduction

The beam modelling for the treatment planning system (TPS) Monaco is done by the provider. The final step in beam modelling is the modelling of multi leaf collimator (MLC) geometry parameters but the provider only does this step in cases where the beam model will be used for VMAT technique. For beam models that will be used for static beam IMRT techniques either Step-and-Shoot or dynamic MLC (dMLC) delivery mode the MLC geometric parameters in the beam models are put to a default value according to the recommendation by the TPS provider. Our experience showed that the default values for Varian linacs are not acceptable for clinical use. The commissioning of the beam model exhibited a dose difference of up to 2% between planned and measured doses.

The provider uses a standard set of test beams for MLC geometry modelling which can be provided to the user for the purpose of beam commissioning [1]. We have observed that the test set isn't optimal for the purpose of testing and optimizing some of the MLC geometry parameters. What is more the TPS provider does not provide any test beams for older models of Varian linacs with older MLC models. As other Monaco TPS users we decided to create our own tests for MLC geometry modelling [2].

This research is focused on the in-house made test beam set and its application in MLC geometry modelling.

Materials and methods

Several test beams were created in Monaco TPS using Monte Carlo calculation algorithm. The fields were shaped by editing the jaw and MLC leaves positions. Due to the limitations of the planning system at creating very specific field shapes the treatment plan files needed to be edited manually by a text editor. Test beams were focused on the following MLC geometry parameters: T jaw transmission, P jaw transmission, leaf transmission, leaf groove width and leaf offset.
**T jaw transmission:**

For each T jaw a separate beam field was created. First an aperture of 26×21,5 cm$^2$ was defined using the MLC leaves. Then one of the jaw was closed over the aperture in such a way that together with the other jaw they formed an aperture of 26×0,3 cm$^2$ which was positioned completely under the upper or lower part of the closed MLC leaves. In this way it was possible to obtain only the transmission through the T jaws.

**P jaw transmission:**

Similarly the two fields for P jaw transmission were created. The MLC aperture of 14×28 cm$^2$ and the jaw aperture of 0,5×28 cm$^2$ were positioned so that one of the P jaws was completely shielding the whole MLC aperture. For the TPS to be able to calculate the beam dose a small intersection of 0,5×1 cm$^2$ between both apertures was created by extending one MLC leaf.

**Leaf transmission**

A two-segment test beam with a jaw aperture of 13×27,5 cm$^2$ was created. In each segment MLC leaves were shielding the whole jaw aperture from one bank or the other with the exception of the top MLC leaf which was needed for the TPS to start the dose calculation.

**Leaf groove width**

A four-segment test beam with a jaw aperture of 13×27 cm$^2$ was created. In the first and the third segments the odd leaves were shielding the jaw aperture from the left and the right bank respectively while in the second and fourth segment the even leaves were used for shielding the jaw aperture in similar way.

**Leaf offset**

Ten dMLC test beams were created with MLC shaped vertical gap of different widths traversing the jaw aperture with constant speed. The gap widths used were 0.2 cm, 0.4 cm, 0.6 cm, 0.8 cm, 1 cm, 2 cm, 3 cm, 5 cm, 7 cm and 10 cm. The width of the jaw aperture was equal to 14 cm plus the MLC gap while the length was 27 cm.

**Dose measurements**

Measurements were done on Matrixx Evolution detector with 1020 ionisation chambers in combination with the MultiCube phantom and OmniPro I'mRT+ 2.0 software [3]. The detector was positioned in the isocentric plane. All the beams were irradiated on Varian DBX single energy linac with the energy of 6MV and gantry, collimator and couch rotations set to 0°. Prior to the measurements, the daily output correction factor was measured using a 10×10 cm$^2$ field. Dose rate of 300 MU/min was used.
Dose calculation

QA plans for all the test beams were created using the CT scan of Matrixx Evolution detector. A non-clinical CT to ED conversion file was created that is only used with plastic phantoms. The relative electron density of the area containing ionization chambers was overridden with the density of the surrounding plastic so that the differences in densities would not perturb the dose distribution in the investigated plane.

Isocentres of all test beams were placed at the centre of the detector at the depth of the effective point of measurement. The dose was calculated with Monte Carlo algorithm with the dose grid of 1mm and the standard deviation between 0,4% and 2,5%. After the calculation the isocentric coronal dose planes for all beams were exported to OmniPro I'mRT+ 2.0 software.

Dose comparison and Parameter optimization

Measured and calculated dose planes were compared using OmniPro I'mRT+ 2.0 software. For every beam a matching region of interest (ROI) was defined on both dose planes and the average dose over each ROI was compared. For the leaf offset parameter an average over all 10 test beams were used.

MLC geometry parameters in the beam model were iteratively tweaked and each time the dose was recalculated and compared to the measurements until a good agreement between both average doses was achieved. First the jaws and MLC transmission parameters were obtained since their tests are designed in such a way that the resulting dose was dependent only on one MLC geometry parameter. Once the values for these parameters were defined leaf groove width and lastly leaf offset parameters were optimized.

Results

The changes in the parameters values were the following: for the T jaw transmission parameter the value changed from 0,0093 to 0,0047 so the transmission through the jaw was reduced by 50%, the P jaw transmission value changed from 0,0093 to 0,0036 resulting in a change of 61%, the leaf transmission value changed from 0,0110 to 0,0140 meaning that the MLC transmission increased by 27%,leaf groove width value changed from 0,5 to 0,3 and finally the leaf offset value changed from 0 to 0,35.

The optimized MLC geometry parameters were put in clinical use and verified by several clinical IMRT plans.

Discussion
A good model of MLC geometry parameters allows planning with a high degree of fluence modulation value changed from 0.0093 beneficial in some extreme cases of IMRT and VMAT treatment planning but even routine cases can benefit from an improved model as the calculated dose represents more accurately the actual received dose.

Additional tests would be needed which would distinguish between MLC transmission and interleaf leakage.

Even with the new linac models we prefer to use our in-house test beam set for MLC modelling rather than the tests provided by the vendor since our tests are designed in such a way that they allow a linear progress through the MLC geometry parameters optimization where the tweaking of the next parameter does not affect the previous one.

References


Comparing planar dose measurement in diagonal and coronal/sagittal planes with Scandidos Delta4 phantoms

Primož Peterlin, Tatjana Pernek, Nevenka Čuk, Attila Šarvari. Institute of oncology Ljubljana, Zaloška c. 2, SI-1000 Ljubljana, Slovenia. ppeterlin@onko.i.si

Introduction

Most radiotherapy centres have adopted a policy that treatment plans for complex radiotherapy techniques like volumetric modulated arc therapy (VMAT) need to undergo pre-treatment verification (Bedford et al, 2009; Korreman et al., 2009). One of the choices for pre-treatment verification are ScandiDos Delta4 phantoms, either the original Delta4PT phantom, introduced in 2006, or the improved Delta4+. Both models employ two orthogonal matrices of 1069 p-Si diodes covering an area of 20×20 cm² embedded in a cylinder of either poly-methyl methacrylate (PMMA) or Plastic Water. They differ, however, in the orientation of the measurement planes: in Delta4PT, they are diagonal, tilted 40° and 50° from the vertical; in Delta4+, they coincide with the coronal and sagittal planes of the common patient setup. In addition, ScandiDos has also made available an adaptor which tilts the Delta4PT phantom so that its planes of measurement coincide with the coronal and the sagittal planes.

An obvious advantage of the measurements in the coronal and the sagittal planes is that they are easier to interpret. In this study, we examined the dosimetric results obtained by three different systems: Delta4PT, Delta4+, and Delta4PT with the coronal-sagittal support (CSS).

Materials and methods

19 VMAT treatment plans for 17 patients on a single treatment machine (Varian NovalisTx; Varian Medical Systems, Palo Alto, USA) were examined. 11 for the pelvic region (mostly prostate or whole pelvic region), 8 for either head-and-neck cancer or brain. Every treatment plan was verified using Delta4PT (ScandiDos, Uppsala, Sweden) phantom before entering the clinical use. In addition, all the plans were verified again three more times: once with Delta4PT phantom, once with Delta4+ (ScandiDos) phantom, and once with Delta4PT phantom using CSS. These measurements were done in a single session each.

Dose deviation, distance to agreement (DTA) and mean gamma index (Low et al., 1998) for individual treatment VMAT fields were analysed. Gamma index analysis with 3% dose deviation and 3 mm positional displacement thresholds was used for treatment plan acceptability verification, with > 95% of points with $\gamma < 1$ indicating a clinically acceptable plan. Delta4 software (release October 2016) was used for the analysis. To minimise the phantom setup variability, phantom position was adjusted using the Optimize phantom position option, optimising the gamma pass rate. We did not use the $\gamma < 1$ pass rate as an evaluation criterion in this study, as with 139 out of 156 treatment fields, all the points fulfilled that criterion, and the
lowest value was 96.8%.

Bland-Altman analysis (Bland and Altman, 1986) was used for testing the agreement of the dosimetric parameter values obtained with different experimental setups.

**Results**

We first tested repeatability of treatment plan evaluation with the Delta4PT phantom in the diagonal setup. 31 treatment fields belonging to 15 treatment plans were compared. The first set of measurements were acquired during a six-month period prior to their clinical use, the second set was acquired during a single session. Figure 1 shows Bland-Altman plots for dose deviation, DTA and mean gamma index ($\gamma$). A single point in the diagram corresponds to a single treatment field, its $x$-coordinate being the arithmetic mean of the values obtained for the chosen dosimetric parameter in both sets, and its $y$-coordinate the difference (eval – ref) of the values in both sets, with the clinical set being the reference set and the single-session set the evaluation set. Dashed line indicates the mean, and dotted lines ±1.96 standard deviation. A one-sample $t$-test confirms that mean value of the differences does not differ significantly from zero in the case of dose deviation pass rate and $\gamma$ ($p = 0.36$ and 0.053, respectively), which is in agreement with our previous findings (Šarvari and Pernek, 2016), while the average DTA pass rate is statistically significantly higher ($p < 0.01$) in the evaluation group.

In the second round of comparisons, 39 treatment fields from 19 treatment plans were compared. The reference set comprised the pre-clinical measurements with the Delta4PT phantom, and the evaluation set the measurements with the Delta4+ phantom acquired in a single session. Figure 2 brings the Bland-Altman plots. In this case, the mean values for dose deviation pass rate, DTA pass rate, and $\gamma$ all differ statistically significant between the two sets. Delta4+ consistently gives lower pass rates for dose deviation and DTA. In addition, we can observe a trend: the agreement is poor for the treatment fields which achieve low scores, and comparable to the previous comparison for the treatment fields which achieve high scores.

In the final round of comparisons, 39 treatment fields from 19 treatment plans were compared. The reference set comprised the pre-clinical measurements with the Delta4PT phantom, and the evaluation set the measurements with the same phantom in the coronal-sagittal setup, acquired in a single session. Figure 3 brings the Bland-Altman plots. A one-sample $t$-test confirms that mean value of the differences does not differ significantly from zero in the case of dose deviation pass rate and $\gamma$ ($p = 0.89$ and 0.84, respectively), while the average DTA pass rate is statistically significantly lower ($p = 0.02$) in the CSS setup.

In the three sessions, the phantom was set on the treatment couch and then a series of treatment plans was irradiated. It was our hypothesis that the setup error is the same for all treatment plans recorded in a single session, and that optimising phantom position would yield the same set of
shifts for all the treatment plans in the session. This did not happen to be the case. In some cases, optimising phantom position did consistently detect a phantom misplacement, yet in other cases, the values obtained were spread over a 2 mm range.

Discussion

The results obtained by the same phantom (Delta4PT) in the diagonal and the coronal-sagittal setup show an agreement which is comparable to the repeatability in the diagonal setup; switching between different detectors (Delta4PT and Delta4+) however does not necessarily yield the same results and should be approached with caution.

As long as $\gamma(3\%, 3\ mm) < 1$ is used as a clinical criterion, none of these differences matter. We believe however that the 3 mm tolerance for the position displacement is too loose. An analysis of the proposed shifts obtained by optimising phantom position can serve as an estimate of the accuracy of phantom positioning, yielding a more realistic value for the position tolerance.

Acknowledgements

The authors thank I. Wiberg of ScandiDos AB for lending us the coronal-sagittal support (CSS) for the Delta4PT phantom.

References


Figure 1: Bland-Altman plots comparing two sets of measurements with the Delta4PT phantom with the detectors in the diagonal planes.

Figure 2: Bland-Altman plots comparing a set of measurements obtained with a Delta4PT phantom in the diagonal setup against a set of measurements obtained with a Delta4+ phantom obtained in the coronal-sagittal planes.
Figure 3: Bland-Altman plots comparing a set of measurements obtained with a Delta4PT phantom in the diagonal setup against a set of measurements obtained with the same phantom (with CSS) obtained in the coronal-sagittal planes.
National participation in an IAEA CRP on quality and accuracy in radiotherapy - preliminary results

Borislava Petrovic1,2, Laza Rutonjski1, Ozren Cudic1, Jelena Stankovic3, Dragomir Paunovic4, Brendan Healy5
1 Oncology Institute Vojvodina, Sremska Kamenica, Serbia
2 Faculty of Sciences, University of Novi Sad, Novi Sad, Serbia
3 Radiotherapy department, Clinical Center Nis, Nis, Serbia
4 Health center Kladovo, Kladovo, Serbia
5 International Atomic Energy Agency, Vienna, Austria

Introduction

In 2013 the IAEA initiated a Coordinated Research Project (CRP) with the aim of investigating the relationship between treatment accuracy and quality assurance extent and depth in radiotherapy. At the national level, surveys of radiotherapy center practices and end-to-end testing according to the guidelines set by the CRP investigators were to be conducted. The results of the CRP will contribute to development of a national QA program in radiotherapy.

Materials and methods

Serbia has 7 radiotherapy centers, where in total 20 linear accelerators are operational. The survey consists of a form to be filled out, and after the third level form, the treatment planning and measurements were to be performed, according to the instructions given by the project team. Three surveys were developed under the CRP, and consisted of a) general questions on radiotherapy facilities, b) questions on staffing and equipment, and c) more detailed questions on types of diseases treated, numbers of patients treated, complexity of treatment techniques and quality assurance practices. The end-to-end test consisted of treatment planning of a given case in an anthropomorphic phantom, which was afterwards treated as if it was a patient, while ionization chamber measurements were performed in defined points of interest.

Results and discussion

Although all centers confirmed their interest, six have responded with first surveys and five with second surveys. The third survey was returned by 3 centers who also performed end-to-end measurements, while 2 did not respond to the third survey due to confidentiality reasons. Consequently, measurements were not performed at those centers.

For the centers who returned the treatment planning and measurement data, good agreement between planned and measured data was registered for 3D-CRT case, as the centers have been audited through national TPS verification under another IAEA project. As for 2D RT cases, good agreement was also registered.
Evaluation of dose planes using 2D detector array—patient specific dosimetry or accelerator performance assessment?

David Rajlić¹, Đeni Smilović Radojićić¹, Božidar Casar², Manda Švabić Kolacio¹, Dario Faj³, Slaven Jurković¹,⁴*

¹University Hospital Rijeka, Medical Physics Department, Krešimirova 42, Rijeka, Croatia
²Institute of Oncology, Department of Radiation Physics, Zaloška 2, Ljubljana, Slovenia
³Faculty of Medicine, University of Osijek, J. Huttlera 4, 31000 Osijek, Croatia
⁴Department of Physics, Faculty of Medicine, University of Rijeka, Braće Branchetta 20, 51000 Rijeka

*Corresponding author: slaven.jurkovic@medri.uniri.hr

Introduction

The accuracy of dose calculation is one of the most important factors in the radiotherapy treatment process. In Monte Carlo (MC) based systems absorbed dose delivered by external photon beam can be reported either as dose-to-media ($D_m$) or dose-to-water ($D_w$). There is a common agreement that MC simulation is the most promising method for accurate calculation of absorbed dose¹,². Nevertheless, there is still no general agreement regarding the choice of the calculation method¹-³. Even so, such algorithms have their own inherent limitations. These limitations are particularly evident in high density media. The differences between $D_m$ and $D_w$ approaches in high-density tissues (e.g. bones) were significant and of opposite sign, when compared to measured values²,³. One of the methods that represent the current standard for patient specific dosimetry is the evaluation of dose planes measured with an ion chamber array inside a homogenous phantom using gamma method⁴,⁵. Considering the fact that the beams pass only through homogenous water equivalent media in this case, this does not replicate the realistic conditions present when a patient is undergoing therapy.

Therefore, to more accurately evaluate the capabilities of the TPS, gamma passing rate was examined for beams passing through inhomogenous phantoms in different geometries in addition to the ones obtained using the water equivalent homogenous phantom using $D_m$ and $D_w$. Also, to better differentiate the underlying reasons for possible discrepancies, a selection of different plans were evaluated, ranging from simple rectangular fields to IMRT plans of different complexity. The results and analysis of this research will be presented in this presentation.

Materials and methods

In this study the research was performed using devices which are in clinical use at Radiotherapy Department of University Hospital Rijeka. Namely, 6MV beam of linear accelerator Siemens Oncor Expression with 160 multileaf collimator (MLC), Siemens Somatom Open CT simulator and Elekta Monaco (v.5.11.) TPS. Linear accelerator was commissioned and prepared for the clinical implementation of IMRT according to international standards.

2D detector array (IBA Matrixx IMRT) with 1020 ion chambers spaced at 0.7 cm distances one from another is used to evaluate TPS accuracy both in homogeneous phantom
(IBA Cube Phantom) and anthropomorphic phantom (CIRS Thorax). Validation was based on gamma analysis with 3%/3mm and 2%/2mm criteria respectively.

For the purpose of a thorough evaluation of the gamma results depending on inhomogeneous media, considering also different complexities of plans, several different phantom configurations in conjunction with the IBA Matrixx detector were used. All phantom combinations were scanned with 0.2cm slice thickness with appropriate relative electron densities tables assigned. In order of increased heterogeneity complexity, the 2D detector was placed under different measuring conditions using homogeneous and semi-anthropomorphic phantoms:

1. MultiCube-IBA MultiCube homogenous phantom
2. PMMA plates (3 cm) and various thicknesses of the CIRS Thorax phantom placed perpendicular to the measuring plane with
   a. CIRS 5N=5 cm thickness
   b. CIRS 10N=10 cm thickness
   c. CIRS 15N=15 cm thickness
3. CIRS Thorax phantom positioned on the detector in regular manner.

All calculations were performed using Elekta Monaco 5.11 TPS with D_w and D_m reporting modes, respectively. In order to achieve an appropriate level of dose calculation accuracy and consistency, all plans were calculated with 0.2cm grid size, 0.5% statistical uncertainty, and „per control point“ calculation mode.

Different QA plans with all beams set to 0° were calculated using all above mentioned phantom configurations, ranging from simple square referent field (15×15 cm^2) to real clinical IMRT plans:

1. IMRT1-prostate plan with 23 segments
2. IMRT2-CNS plan with 40 segments
3. IMRT3-H&N plan with 76 segments
4. IMRT4-H&N plan with 3 dose levels integrated boost (105 segments)

Results and discussion

The gamma analysis results for different measuring geometries and different levels of plan complexity as well as different reporting modes are presented in Table 1. and Figures 1 and 2. The results for IMRT plans are degrading depending on the thickness of non-water equivalent material of up to 9% for 3%/3mm, regardless of reporting mode- D_m or D_w. Additionally, the gamma passing rates were degrading depending on the level of complexity of plans, up to 15% when looking at 2%/2mm.

These results together with the fact that 2%/2mm passing rates degrade more rapidly than 3%/3mm suggest that the resolution of the detector is one of the limiting factors of the analysis. Additionally, there is a significant difference in the passing rates depending on D_w and D_m reporting modes.

Our results raised question of possible limits of the gamma method in assessment of plan delivery quality. Consequently, good results obtained using standard patient specific dosimetry methodology does not guarantee the accuracy of delivered dose distribution in real clinical cases.

Reference

<table>
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<tr>
<th>Phantom</th>
<th>Gamma/%</th>
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<th>IMRT 2</th>
<th>IMRT 3</th>
<th>IMRT 4</th>
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<td>86,14</td>
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<td>2%/2mm</td>
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Table 1. Results of gamma analysis for different measuring geometries, different levels of plan complexity and different reporting modes.
Figure 1. Gamma passing rates for dose-to-water reporting mode depending on different measuring geometries and different levels of plan complexity.

Figure 2. Gamma passing rates for dose-to-media reporting mode depending on different measuring geometries and different levels of plan complexity.
COMPARISON OF MANUAL AND APPLICATOR LIBRARY RECONSTRUCTIONS FOR SEGMENTED VAGINAL CYLINDERS

Tamara Jovanovic  
Clinical Center Nis, Clinic of Oncology, Nis Serbia  
tamaradjj@gmail.com

Introduction  
Brachytherapy (BT) is a very local treatment of tumors that uses radioactive sealed sources which are placed near or inside the site to be treated. A dose fall-off around the source has a very steep gradient and is highly conformal. BT is applicable either with temporary or permanent implants. High dose rate (HDR) BT with temporary implants uses miniaturized stepping source of Ir-192. It is the most common and safety way of the dose delivery that assumes remote afterloading technology.

The dose distribution are non-homogenous within the treated volume of the BT implant\(^{(1)}\). Because of steep dose gradients there is a sleeve of high dose surrounding each source. With the stepping source afterloading machines, both the dwell positions and the dwell times can be adjusted to make the dose distribution as homogenous as possible throughout implant. This optimization method allows doses to normal tissue to be minimized.

BT treatment planning system (TPS) has a few methods of dose optimization (CT). Computed tomography (CT) is “state of the art” nowadays for BT planning. The most versions of TPSs have possibility to use pre-defined models of standard applicators for reconstruction. These are models of rigid applicators with a source path defined, and there are confined in the virtual \(^{(2)}\) applicator library. This paper investigates influence on dose distributions around segmented vaginal cylinders when compared manual and applicator library reconstructions.

Materials and methods  
When planning directly from CT scans there is much better insight to the dose distribution. In the radiotherapy department in Nis, Serbia, since 2006 CT-based BT planning has been performed for gynecological (GYN) carcinomas of cervix and endometrium with a commercial TPS (BrachyVision, VarianMedical Systems, Inc. Palo Alto, CA). An each fraction for patients undergoing BT is completely individualized. Now with the version 13.6 it is possible to use the virtual BT applicator library and reconstruct standard applicators, such as segmented vaginal cylinders, with the “insert new solid applicator“ option (figure 1). In this study, for five patients on postoperative GYN BT an alternate plans with manual reconstruction were done. The lumen of the catheter is well visualized in CT images. This means that a markers is not necessary to use because of artifacts in the CT images.

All patients had three BT fractions with 6 Gy or 7 Gy per fraction with segmented vaginal cylinder of diameter 3 cm and stump (S) front segment. This cylinder is the most often used and the dose is prescribed on 5 mm from its surface. For each fraction both plans were done with the applicator reconstruction by importing library model and manually by visualizing the catheter lumen. The distance from the applicator tip to the catheter lumen is known and for all diameters of segmented vaginal cylinders with S front segment has value 3,43 mm. Planning were performed with the same window and level of 400 HU. There were defined the dose reference points, bladder and rectum reference points.

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Results

For five patients 15 plans were analyzed and compared for each reconstruction method. Conventionally the dose is defined at 5 mm from the surface of the applicator. Treated volume includes the vaginal vault and the upper part of the vagina. The prescription doses per fraction are 6 Gy or 7 Gy, while treatment lengths 3 cm to 6 cm. Appropriate dwell times are put with geometrical optimization, then adjusted with the dose shaper and renormalized. Graphical optimization refers to a manual graphical dose shaping technique to improve the dose distribution. There were generated a reference lines at distance 5 mm away from the cylinder surface, a reference line 5 mm from the tip of the applicator and, with manual reconstruction, also an applicator tip.

Following the ICRU Report 38 recommendations\(^{(3)}\), there were compared total reference air kerma (TRAK), volume receiving 100% prescription dose (V100\%) volume receiving 200% prescription dose (V200\%) and prescription points doses. The values of applicator library reconstruction plan is normalized with the values of manually reconstruction plan (table 1).

For the difference between both reconstruction, the average absolute differences as a percentage were 0.3% for TRAK, 0.01% for V100\% and negligible for V200\%.

Discussion

The use of CT imaging for HDR BT treatment planning is increasing, particularly for the management of GYN malignancies. A virtual applicator library for applicator reconstruction with 3D image has been available for most HDR BT TPSs\(^{(2)}\).

Four parameters were compared between two reconstruction methods. A significant difference between applicator library and manually reconstruction methods does not exist. An applicator library reconstruction allows easier image-based BT planning.

References

Figure 4 The dose distribution around the solid segmented vaginal cylinder.

<table>
<thead>
<tr>
<th>Fr.No.</th>
<th>I</th>
<th></th>
<th>II</th>
<th></th>
<th>III</th>
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<td>3.</td>
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<tr>
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<td>1.02</td>
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<td>0.99</td>
<td>1.02</td>
<td>1.01</td>
<td>1.01</td>
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</tr>
</tbody>
</table>

Table 1 Normalized values of four parameters for applicator library reconstruction.
LDR versus HDR prostate brachytherapy: dosimetric comparison of intraoperative plans

Zsanett Bianco-Molnár¹, Péter Ágoston², Kliton Jorgó², Tibor Major²

¹Budapest University of Technology and Economics, Budapest, Hungary
²National Institute of Oncology, Budapest, Hungary

E-mail: zsanetmolnar@yahoo.com

Introduction

The prostate cancer is the fourth most common cancer in both sexes combined and the second most common cancer in men [1]. Permanent low-dose-rate (LDR-BT) and temporary high-dose-rate (HDR-BT) brachytherapy are competitive techniques for clinically localized prostate radiotherapy [2]. The aim of the work is to dosimetrically compare the permanent LDR-BT and HDR-BT as monotherapy [2, 3] for patient with prostate cancer.

Materials and methods

For the investigation 25-25 low- and medium-risk patients with prostate cancer treated with LDR and HDR technique were selected. LDR-BT was performed with permanent I-125 isotopes (“seeds”) implanted in the prostate. The stranded seeds were separated 1-1 cm from each other. The prescribed dose was 145 Gy. HDR-BT technique was performed with Ir-192 source using an afterloading equipment. The prescribed dose for the prostate was 19 Gy in one fraction. In both methods transrectal ultrasound (UH) imaging was used with a template for needle guidance. On the basis of preimplant plans needles were inserted in the prostate, then with live UH image-guidance the true positions of the needles were updated in the final intraoperative plans. For the dosimetric comparison we prepared LDR plans to the existing HDR plans and vice versa. For both techniques Oncentra Prostate v4.3 (Veenendaal, Netherlands, Elekta Brachytherapy) planning system with inverse dose optimization algorithms was used. In the investigation relative doses were calculated and compared because the prescribed doses were very different. The target coverage with V100, the homogeneity with DHI, and the conformity with COIN indices were described. To characterize the dose to the urethra we used $D_{\text{max}}$, $D_{10}$, $D_{30}$, $D_{0.01cm^3}$ and $D_{0.1cm^3}$ parameters. For the evaluation of the rectum we used $D_{\text{max}}$, $D_{0.1cm^3}$, $D_{1cm^3}$, and $D_{2cm^3}$ parameters [4, 5]. The statistical comparisons were made by Wilcoxon signed-rank test with $p < 0.05$ significance level.

Results
The average size of the target volume was 37.4 cm$^3$, the mean volume covered by reference isodose was 48.1 cm$^3$. The V100 was nearly equal in the two techniques, in LDR it was 98% and 97% in HDR. The V150 and V200 were significantly smaller in the HDR (32% vs. 59% and 10% vs. 24% respectively). The HDR technique resulted in more homogenous dose distributions (DHI: 0.67 vs. 0.41). However, the D90 was higher in the LDR (120% vs. 110%). HDR dose distribution was more conformal (COIN: 0.78 vs. 0.70). These differences were significant except for the V90. All investigated parameters for the urethra were significantly lower in HDR than in LDR. D$_{max}$: 123% vs. 156%, D10: 117% vs. 135%, D30: 113% vs. 128%, D0.01cm$^3$: 120% vs. 146%, and D0.1cm$^3$: 118% vs. 137%. The results were similar to the dose to the rectum. D$_{max}$: 84% vs. 112%, D0.1cm$^3$: 78% vs. 93%, D1cm$^3$: 65% vs. 73% and D2cm$^3$: 58% vs. 63%.

Discussion

There are no significant differences in dose distribution of target volume between the two types of technique. In HDR plans the dose distribution was more homogenous and conformal than in the LDR plans. The dose to the OARs was lower in the HDR plans. The validation of the results requires longer follow-up with clinical data assessment.

References

Abstract: Radiotherapy (RT) is a multidisciplinary modality developed according to the progress in physics and technology. Simulators are medical devices used in radiation oncology clinics to simulate external beam treatments. All conformal external beam techniques are, in a broader sense, image guided, and consequently allow better treatment curability. Computed tomography (CT) is primary among all imaging modalities and it is the method of choice for a three-dimensional therapy planning (3D). An alternative device to the classical simulator that uses CT with a system-software for treatment simulations is a virtual simulator (VSim). It is the CT simulator that performs the simulation process using the patient’s anatomical model. In this paper, a dedicated CT scanner Siemens Sensation Open with the Coherence Dosimetrist software would be presented throughout the virtual simulation protocol, and accordingly the 3D approach in therapy simulation would be illustrated.

Keywords: treatment simulation, conformal techniques, computed tomography, virtual simulation.

Introduction
Radiation therapy uses high-energy photon rays in order to deliver a very accurate dose of radiation to a well-defined target volume with sparing surrounding healthy tissues. The desired goals are eradication of the tumor and the improvement or prolonging of patient’s life. RT is a very demanding process, that requires accuracy and effectivity, composed of several steps. The essential step in external beam therapy planning is very clear and precise definition of the tumor location and boundaries from all available data (clinical, surgical, radiological and pathological). The second important step in this process is the therapy simulation. This procedure implies localization of the target and surrounding healthy structures in relation to the therapy machine, and accordingly definition and verification of radiation fields.

During the eighties of the twentieth century, special CT scanners, named virtual simulators, are designed for needs of RT. Implementation of virtual simulators in the clinical practice was a very significant advance in radiation oncology. This concept of CT simulation (VSim), mimics the simulation process for the conventional therapy and allows 3D simulation. The virtual simulation is a part of the treatment preparation and planning based on the computer definition of patient’s model, that does not require his presence. We introduced 2011. the virtual simulator Siemens Sensation Open (Siemens, Germany) with a Coherence Dosimetrist software (Siemens Medical Solutions, USA) into the clinical practice. It is dedicated for radiotherapy with large bore opening of 82 cm, flat-top couch and external movable lasers (figure 1).

Materials and methods
Tomography means cross-sectional imaging of any part of the human body, while improving a soft tissue contrast and providing an information about structures in space. CT is a technology that allows the non-destructive evaluation of the internal structure of the human body. The basics of CT imaging are that of X-ray principles, when the film is substituted by a
detector, which measures the X-rays profile. The CT scanner consists of a gantry, which includes the X-ray source, X-ray detectors, and data acquisition system, a patient table, a control console and a computer. With the progress of computer technologies it was possible to assure 3D CT simulation. Virtual simulators are CT scanners with a software option for radiotherapy, a large bore opening for immobilization, a flat-top couch compatible with the treatment table-top and a system of movable external lasers for patients positioning. In our radiotherapy department, VSim concept consists of:

1. an isocenter placement in the center of gross tumor volume (GTV) or clinical target volume (CTV).
2. a virtual definition of radiation beams in relation to target volumes and organs at risk in absence of the patient.

The virtual simulation assumes positioning and immobilization of the patient, CT acquisition, Dicom import of CT series into the VSim software and accurate determination of the beams isocenter using patient’s volume images. We adopted the first method for performing this procedure that requires the oncologist to be present to identify the target volume and the isocenter from the scan information (volume images displayed in multiple image planes). While the patient remains in the treatment position, isocenter coordinates are sent to the external lasers system and “marked” on the patient’s skin. Then the patient goes and his virtual anatomical model is used for treatment planning. Other parts of VSim process are concerned with our workflow that assumes either 3D conformal technique or virtual therapy simulation.

**Results**

The effect of RT treatments depends on the precise delivery of high irradiation dose on the tumor site with sparing surrounding healthy tissues. Therefore a patient positioning, a target volume definition and a irradiation field placement are crucial steps while planning the irradiation process. Briefly in the our current clinical protocol for the virtual simulation, the patient goes through the following steps:

- Patient positioning and immobilization on the flat-top couch of a CT scanner in the treatment position. Alignment of the patient is made with lateral wall laser and sagittal laser. Opaque catheters don’t have to be used as visual markers.
- CT acquisition according to the particular protocol for the site to be treated.
- Network import of CT images into the VSim application.
- Delineation of an external contour of the patient (skin).
- Localization of the target structure GTV (CTV) with an interpolation method. The radiation oncologist performs quickly delineation using a few CT slices from the beginning, the middle and the end of the structure and then uses “interpolate” option. A beam isocenter is defined as a reference point that is the center of the structure (figure 2).
- Transfer isocenter coordinates to the software which enables moving of external lasers (“marked patient” option).
- The patient is “marked” where the laser projection illuminates the skin and finally the patient is removed from the couch.
- Definition of an arbitrary virtual beam due to the transfer of isocenter coordinates into the treatment planning system (TPS) (figure 3).
- Target volumes and organs at risk delineation or virtual beams placement. In the latter case, the isocenter is placed in the center of region of interest using “fluoro mod” option (figure 4).
- Dicom RT transfer into the Focal Sim (CMS; Germany) application, for the reason that a direct interface to the treatment planning system does not exist.

Discussion
The advantages of CT imaging in radiation oncology has been recognized through a better staging of the primary tumor, a visualization of anatomical details in the transverse plane (CT localization) and assisting in a computerized therapy planning. The advantages of CT-based virtual simulation are well known and include the fact that target volumes, critical organs and structures can be effectively defined and displayed in multiple image planes (axial, coronal, sagittal or oblique). In VSim it is possible to display on the same screen (a) the beam’s eye-view, where the Digital Reconstructed Radiograph (DRR) is displayed, (b) the room view including a 3D model of the simulator or the treatment machine and (c) the observer’s eye-view, where the 3D surface reconstruction of the patient is shown (figures 3 and 5). These images offer the user an overview of the simulation and treatment planning process. Also, it is possible to fuse with others imaging modalities (figure 5).

Inherent in all successful CT Sim techniques is the appropriate immobilization of the patient that is compatible with the constraints of the CT scanner. The VSim process depends on defining a relationship between the CT image coordinates (patient) and the treatment coordinates (machine) that allows a precise transformation from the localization setup to radiotherapy treatment coordinate space. Verification takes place only on the treatment unit with the electronic portal imaging system if the isocenter coordinates are determined.

In our clinical practice, there were problems with issues related to implementation of the virtual simulation protocol, very demanding immobilization equipment such as belly board, wrong tattoos on the patient’s skin and invalidation of isocenter coordinates in the FocalSim software. Nowadays, Vsim concept has become an example of the successful clinical practice.

The principal goals of virtual simulation are: a) to position the patient in order to find an optimal 3D treatment plan for each patient by manipulating a virtual patient model and evaluating of multiple options before treatment; b) to identify the isocenters or other reference points on the patient’s skin.

References
4. Zimeras, S. Virtual simulation for radiation therapy treatment using CT medical data.
Figure 5 The virtual simulator dedicated for RT.

Figure 2 GTV delineation using an “interpolate” option and the isocenter placement in its center.
Figure 3 An arbitrary virtual beam placement with ‘marked’ isocenter.

Figure 4 A ‘fluoro mod’ option for the virtual therapy simulation.

Figure 5 Multiple image planes with MRI fusion
BNCT AS RADIOSENSITIZER IN HIGH-ENERGY RADIOTHERAPY TREATMENTS

Katia Alikaniotis¹, Gianrossano Giannini¹, Alba Zanini², Silvia Anglesio³

¹University of Trieste & INFN sec. Trieste, Via A. Valerio 2, 34127, Trieste, Italy
³INFN sec. Torino, Via P. Giuria 1, 10126, Torino, Italy
⁴Hospital “San Luigi Gonzaga”, Regione Gonzole 10, 10043 Orbassano (TO), Italy

e-mail: katia.alikaniotis@gmail.com

Introduction

Present-day cancer treatments still require further improvements in order to obtain a better dose control to target volume, reducing the incidence of secondary radio-induced tumors. One of the main drawbacks when dealing with radiotherapy is the necessity to precisely select cells to be treated, reducing the damage to the healthy ones. Today the efficiency of newer radiosensitizers, acting on tumor cells, is investigated in many tumor diseases to improve the radiotherapy effectiveness.

One of the techniques of high-selectivity radiotherapy is BNCT (Boron Neutron Capture Therapy). This therapy is based on neutron capture by 10-boron (¹⁰B), and it is a selective therapy because the ¹⁰B transporting carrier is preferentially accumulated in tumor cells due to their faster metabolism. A ¹⁰B compound (usually ¹⁰B-Phenyl-Alanyne - ¹⁰BPA) is administered to the patient, since ¹⁰B has a high thermal neutron (E < 0.4 eV) capture reaction cross section (3843 barns at 0.025 eV). During neutron irradiation, a nuclear reaction takes place producing heavy fragments from ¹⁰B, an α particle and a 7-lithium (⁷Li) nucleus. The two generated particles have a high-LET and a short range (of the order of few µm, consistent with the cell size), thereby selectivity affecting only tumor cells during neutron irradiation.

During X-rays radiotherapy treatments (RT) using electron linear accelerators (e-LINAC) with energies E ≥ 10 MV, patients undergo to an undesirable neutron dose [1]. Neutron
production results from the interaction of high-energy photons with various high Z nuclei present in e-LINAC gantry by photonuclear reaction (γ,n). The production is governed by the Giant Dipole Resonance reaction (GDR) and neutrons are generated when the incident photon energy exceeds the GDR reaction threshold (6 MeV - 20 MeV), with a mean energy of about 1 MeV and an isotropic angular distribution. The elastic scattering on light elements constituting the human body moderates this undesired neutron component. Therefore, a consistent thermal neutron flux (of about \(10^7 n_{th} \text{ cm}^{-2} \text{ Gy}^{-1}\)) useful for BNCT application is localized in the tumor area.

The study analyses the possibility to employ this thermal neutron background for BNCT applications in order to enhance the radiotherapy effectiveness. The previous work proves that the thermal neutron peak could be exploited for BNCT, delivering to the patient an additional therapeutic dose of about 4% (or more) to the photon dose [2]. It’s now necessary to verify if this BNCT additional dose is more concentrated in the tumor tissue and much less in healthy ones: two typical radiotherapy sessions of two patients affected by prostate cancer and lumbar vertebra cancer have been so studied.

**Materials and Methods**

A simplified tissue equivalent anthropomorphic phantom has been exposed to an 18 MV photon beam in order to evaluate the BNCT effect in two real case cancer treatments (prostate and lumbar vertebra) by means of bubble dosimeters for thermal neutrons (BDT) placed inside the phantom in suitable holes corresponding to critical organs.

The anthropomorphic phantom, Jimmy, has been designed and built by INFN of Turin in collaboration with the Ispra JRC (Join Research Centre), Varese-Italy. It was specifically developed for neutron dosimetry in order to evaluate the neutron equivalent dose in tissue. Different slabs of plexiglass and polyethylene make up the phantom, and a human bone dust is inserted in it in correspondence of the vertebral column. There are also 16 cavities in correspondence of critical organs suitable to locate integral passive bubble dosimeters (BTI Bubble Tech. Ind., Ontario, Canada)[3]: BD-PND sensitive to fast neutrons (100 keV < \(E < 20\) MeV) and/or BDT for thermal neutrons (\(E < 0.4\) eV); accuracy BTI: ± 20%. Jimmy has been designed following the International Commission on Radiological Protection indications for neutron dosimetric phantoms both about organ positions (ICRP 60) and tissue substitutes (ICRU 44). The phantom external dimensions are: head (15×13.5×19) cm³; neck (10×11×13.5) cm³; body (59×(30-36)×20) cm³. The weight is about 37 kg.
The anthropomorphic phantom Jimmy and BTI bubble dosimeters are shown in Fig 1.

In this work, the anthropomorphic phantom Jimmy has been considered as a real patient: firstly, it performed a CT (Computered Tomography) and the corresponding images were analyzed for the treatment planning by using the “Oncentra Masterplan 4.3” software. Secondly, Jimmy has undergone the radiotherapy session with bubble dosimeters placed inside it in order to measure the thermal neutron dose during the radiation exposure.

For the real prostate cancer treatment taken in consideration, radiation exposure characteristics are: 18 MV e-LINAC ELEKTA PRECISE in photon mode; field (10x10) cm² at isocenter; five AP (anterior-posterior) fields at 0°, 45° (wedge 60°), 90°, 270°, 315° (wedge 60°); prescript dose at isocenter 2 Gy per session; total dose 70 Gy.

For the real lumbar vertebra treatment, radiation exposure characteristics are: 18 MV e-LINAC ELEKTA PRECISE in photon mode; field (10x10) cm² at isocenter; two PA (posterior-anterior) fields at 140° (wedge 60°) and at 220° (wedge 60°); prescript dose at isocenter 8 Gy per session; total dose 16 Gy.

To evaluate the BNCT additional dose to the photon dose, it’s necessary to calculate the BNCT weighted biological dose $D_w$, expressed in terms of photon-equivalent unit (Gy-eq or Sv). It takes into account the various physical dose components, arising from neutrons and gamma interactions with biological tissues and $^{10}$B captured by cells:

$$D_w = w_\gamma D_\gamma + w_n (D_H + D_N) + w_B D_B$$

where gamma dose ($D_\gamma$), fast neutron dose ($D_H$), thermal neutron dose ($D_N$) and boron dose ($D_B$) are the physical dose components, while $w_\gamma$, $w_n$ and $w_B$ are their weighting factors. The weighting factors values are respectively $w_\gamma = 1$ for photons, $w_n = 3.2$ for neutrons, $w_B = 1.3$ for boron in healthy tissue and $w_B = 3.8$ for boron in tumor tissue.

Results

The equivalent neutron dose absorbed by the phantom during the radiotherapy session is shown in Fig 2 for the prostate RT and in Fig 3 for the lumbar vertebra RT.
Notice that all results are always normalized to 1 Gy of photon dose absorbed by the phantom at build-up.

Thanks to dose-to-fluence conversion factors tabulated in NCRP 38 [4], it was possible to obtain the thermal neutron fluence in the tumor area: $2.88 \times 10^7 \text{n}_{\text{th}} \text{cm}^{-2} \text{per Gy}$ for the prostate RT, $2.16 \times 10^7 \text{n}_{\text{th}} \text{cm}^{-2} \text{per Gy}$ for the lumbar vertebra RT. A consistent thermal neutron flux is so present in the target volume: the $^{10}\text{BPA}$ administration to the patient has been so simulated by using the MCNP4B-GN Monte Carlo code and the boron dose ($D_B$) has been evaluated. The BNCT weighted biological dose ($D_w$) difference between normal tissue and tumor tissue is only due to the boron component $D_B$; the other BNCT physical components are not here considered because they are always present, even without $^{10}\text{BPA}$ perfusion.

In Fig 4 and in Fig 5 the equivalent boron dose due to $^{10}\text{BPA}$ administration to the patient is reported for the prostate treatment and lumbar vertebra treatment respectively.

In Table 1 all results for both the studied real cancer treatments are summarized.

**Discussion**

In this work the possibility to perform a coupled treatment with high-energy e-LINAC and BNCT is examined. A patient undergoing a conventional radiotherapy treatment is always affected by an undesired neutron dose, which presents an intense thermal neutron flux (of about $10^7 \text{n}_{\text{th}} \text{cm}^{-2} \text{per Gy}$) localized in the target volume. If a boron compound ($^{10}\text{BPA}$) is previously administered to the patient, this neutron component is expected to produce a localized BNCT effect, with a localized therapeutic dose enhancement following tumour characteristics: the BNCT effect is mainly present in the tumour tissue, while nearby organs are preserved.

This application is a preliminary study of the possibility to exploit the undesirable neutron contamination for the enhancement of radiotherapy treatments, especially considering the new trend in radiotherapy consisting of dose escalation and dose hypo-fractionation.

A definitive indication of the coupled treatment using both high-energy e-LINAC and BNCT could be obtained through a biological study of different cell lines, tumor and healthy cells, exposed to X-rays radiation with and without $^{10}\text{BPA}$ perfusion: experimental measurements on human cell lines (bronchial epithelial cells BEAS-2B and lung adenocarcinoma cells A549) are so in progress.

**References**


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Fig 1.
The anthropomorphic phantom Jimmy and BTI bubble dosimeters for thermal neutrons (BDT).
On the left image, it’s also possible to see BDTs placed inside the phantom in suitable holes corresponding to critical organs.

Fig 2.
Thermal neutron equivalent dose at organs for the prostate cancer case studied. BDT measurement.

Fig 3.
Thermal neutron equivalent dose at organs for the lumbar vertebra cancer case studied. BDT measurement.
Fig 4.
Equivalent boron dose due to BPA administration for a thermal neutron fluence in the treatment area of $2.88 \times 10^7$ n cm$^{-2}$ per Gy. Healthy to tumor tissue $^{10}$B ratio 1:3. MCNP4B-GN simulation result. Boron concentration in bladder, the sick organ, is about one order of magnitude higher than nearby healthy organs.

Fig 5.
Equivalent boron dose due to BPA administration for a thermal neutron fluence in the treatment area of $2.16 \times 10^7$ n cm$^{-2}$ per Gy. Healthy to tumor tissue $^{10}$B ratio 1:3. MCNP4B-GN simulation result. Boron concentration in the low column, the sick organ, is about one order of magnitude higher than nearby healthy organs: healthy tissues are preserved by the BNCT effect.
Table 1.

Results of the two studied real cancer cases. During both treatments, patient undergoes to a consistent, undesired and unavoidable neutron component at organs. The thermal neutron component could be exploited for BNCT application, delivering

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<th>Radiotherapy treatment at lumbar vertebra</th>
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<td>5 AP</td>
<td>2 PA</td>
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<td>2 Gy</td>
<td>8 Gy</td>
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<td>70 Gy</td>
<td>16 Gy</td>
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<td>Thermal neutron fluence in the treatment area</td>
<td>2.88E07 cm⁻² Gy⁻¹</td>
<td>2.16E07 cm⁻² Gy⁻¹</td>
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<td>Total thermal neutron dose tumor area</td>
<td>21.0 mSv</td>
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