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FOREWORD

Due to the rapid technological development in the world today, the role of physics in modern medicine is of great importance. The frequent use of equipment that produces ionizing radiation further increases the need for radiation protection, complicated equipment requires technical support, the diagnostic and therapeutic methods impose the highest professionals in the field of medical physics. Thus, medical physics and biomedical engineering have become an inseparable part of everyday medical practice.

There are a certain number of highly qualified and dedicated professionals in medical physics in Macedonia who committed themselves to work towards resolving medical physics issues. In 2000 they established the first and still only professional Association for Medical Physics and Biomedical Engineering (AMPBE) in Macedonia; a one competent to cope with problems in the fields of medicine, which applies methods of physics and biomedical engineering to medical procedures in order to develop tools essential to the physicians that will ultimately lead to improve the quality of medical practice in general.

The First National Conference on Medical Physics and Biomedical Engineering was organized by the AMPBE in 2007. The idea was to gather all the professionals working in medical physics and biomedical engineering in one place in order to present their work and increase the collaboration among them. Other involved professions such as medical doctors, radiation technologists, engineers and professors of physics at the University also took part and contributed to the success of the conference. As a result, the Proceedings were published in Macedonian, with summaries in English. In order to further promote the medical physics amongst the scientific community in Macedonia, our society decided to organize The Second Conference on Medical Physics and Biomedical Engineering in November 2010. Unlike the first, this one was with international participation. This was very suitable event for our Association’s 10th anniversary. We had the honor of hosting some of the most outstanding lecturers, experts in the field of medical physics and biomedical
Lecturers, who consistently participate in high level organized symposia, conferences, and congresses, evaluated our conference as a highly successful event.

The success of the previous conference encouraged us to continue the tradition of the three-year evaluation of our work and organize the THIRD CONFERENCE ON MEDICAL PHYSICS AND BIOMEDICAL ENGINEERING in October 2013. This event is organized again thanks to the cooperation of the European Federation of Organizations for Medical Physics (EFOMP). Renowned experts took part as Conference’s plenary speakers. This proceeding contains works in different fields presented on the conference orally or as poster presentations.

I want to express my special thanks to the invited lecturers for their immense contribution to the success of the conference.

My sincere appreciation to the Scientific Committee for their professional work, and I heartily thank all members of the Organizing Committee for their thoughtful participation, constructive remarks and invaluable assistance.

For closing, I wish to take this opportunity to thank all the participants, co-organizers of this event for their contribution. This scientific event would not be possible without the financial support of the selected companies presented here.

As a chairman of the conference Organizing Committee, I would like to personally thank all the individuals and institutions whose co-creative efforts and support help us to make that kind of conferences possible in Macedonia. This has been a most educational, rewarding, and worthwhile journey.

On behalf of the organizing committee of Third MPBE conference,

Sonja Petkovska
President of the Association for Medical Physics and Biomedical Engineering
DOSIMETRY EXPERIENCE OF 192IR SOURCES USED IN HDR BRACHYTHERAPY FOR CERVICAL CANCER.

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Abstract – Purpose/Objective: The 192IR Sources are the most commonly used in radiotherapy treatments worldwide. According to international recommendations on quality assurance in HDR-brachytherapy, an acceptance test based on the determination of the source strength of any new source shall be carried out before first application to verify the manufacturer’s calibration data. The present paper gives the experimental determination of the source strength for our brachytherapy sources used until now in brachytherapy treatments.

Materials/Methods: At Mother Teresa University Hospital we have a cost-effective gynecological brachytherapy unit from Eckert & Ziegler BEBIG named GyneSource® that is a five channel HDR afterloader equipped with an 192Ir source. The software used is HDRplus™ 2.5 that delivers an optimized treatment plan and makes the process especially fast and we use intracavitary BEbig applicators. From April 2009 up to December 2012, we have imported nine HDR 192Ir Sources. The exchange of the source and acceptance test is done by the physicist of the clinic once the source is imported. The measurements are done with a Well-type ionization chamber HDR1000 Plus and the electrometer used is MAX4000. Only seven sources are compared as we miss the dosimetry data of the first source, and the forth source was not measured and not used because the machine was not working in that time.

Results/Conclusions: Eight sources were accepted for clinically use as the measurement were within the tolerance. The source number four with a deviation of -1.92% has been double checked compared with a free in-air measurement with farmer type chamber that gave a deviation to source certificate of 4% that is still inside the tolerance to accept a source for clinical use. The deviations of measured Air Kerma rate to the value of the sources certificates of all our used 192IR sources are less than 2%, which are within the tolerance. The checked value of updated source strength in planning system and afterloading units for all sources was done and was correctly calculated. A verification with other chambers has been done and also has given very good results. An IAEA external audit has acknowledged the good experience.

Keywords – Air Kerma Rate, 192Ir Source, Brachytherapy, Source verification

1. INTRODUCTION

The HDR sources of 192Ir has a wide range of use in brachytherapy applications through the applicators or needles that permits to place the source near to the tumor in different anatomical locations: cervix, prostate esophagus etc. The dosimetric characteristics of this sources is fundamental for the treatment planning process in brachytherapy. According to international recommendations on quality assurance in HDR-brachytherapy, an acceptance test based on the determination of the source strength of any new source shall be carried out before first application to verify the manufacturer’s calibration data.

At Mother Teresa University Hospital we have a cost-effective gynecological brachytherapy unit from Eckert & Ziegler BEBIG named GyneSource® that is a five channel HDR afterloader equipped with an Ir-192 source. The software used is HDRplus™ 2.5 that delivers an optimized treatment plan and makes the process especially fast and we use intracavitary BEbig applicators. From April 2009 up to December 2012, we have imported nine HDR 192Ir Sources. The number of patients treated during this period was 188. The machine has been working during this four years only 58% of the time, due to different reasons like technical problems and lack of maintenance contracts, lack of qualified personnel the first year of use and now lack of sources due to economical reasons. The unit was bought through IAEA projects.
1.1 Air-Kerma Strength \( S_k \)

Air-kerma strength is a measure of brachytherapy source strength, which is specified in terms of air-kerma rate at a point along the transverse axis of the source in free space. It is defined as the product of air-kerma rate at a calibration distance, \( d \), in free space,

\[
K_{\text{air}}(d),
\]

measured along the transverse bisector of the source, and the square of the distance, \( d \):

\[
S_k = K_{\text{air}}(d) \times d^2 \left[ \mu\text{Gy} m^2 h^{-1} \right]
\]

(1)

The calibration distance must be large enough that the source may be treated as a mathematical point. In practice, the air-kerma rate standardization measurements are performed in air and corrections for air attenuation are applied if needed. Since the measurements for source-strength calibration may be performed at any large distance, \( d \), the air-kerma rate is normally specified in terms of a reference calibration distance, \( d_0 \), which is usually chosen to be 1 m.

\[
\mu U = 1 \mu\text{Gy} m^2 h^{-1}
\]

(2)

The unit of \( S_k \) is denoted by the symbol \( \mu U \) and the protocol TG-43 recommends to fix the distance \( d \) at 1 m in the direction perpendicular to the source axis, \( \theta = \pi/2 \), when the linear sources Air Kerma Strength is determined.

For clinical dosimetry, brachytherapy sources are characterized by their relative dose rate distribution in an unbound water phantom. Absolute values will be achieved by multiplying the relative data with the reference absorbed dose rate, i.e. the absorbed dose rate to water at the source reference point which is defined at 1 cm distance vertical to the source axis. The reason for using the physical quantity air-kerma rate to specify source strength instead of absorbed dose rate to water is the present absence of an appropriate primary standard for this application.

1.2 \(^{192}\text{Ir} \) source Bebig model and design

The \(^{192}\text{Ir} \) source design type produced by Eckert & Ziegler that we use is Ir2.A85-2. The radioactive material is inclosed by a cylindrical high qulity steel capsule which is closed in a seal tight condition by means of laser welded cover. The metallic iridium is presented in the form of 3.5 mm long wire (diameter 0.6mm). The capsule is welded in the bottom side with a flexible cable on whose cylindirical end piece the source designation is located.

\[
\mu U = 1 \mu\text{Gy} m^2 h^{-1}
\]

The experimental quantity \( K_{\text{air}}(d) \) is the product of the measurement \( M \), represented by the current of the electrometer corrected to the reference conditions, and the corresponding calibration factor \( N_k \) determined by the secondary standard dosimetry laboratory (SSDL) for a given source type.

\[
\mu U = 1 \mu\text{Gy} m^2 h^{-1}
\]

The current \( M \) represents the maximum value of the response curve measured by stepping the source along the central chamber axis. The sensitivity of the chamber versus the source position along the guide tube has been checked by varying the position of the source along the length of the guide tube inside the measuring volume. The maximum reading was given in almost all the measurements at the middle of the chamber evaluated with different steps of dwell positions. That has guaranteed the same position characteristics of the well-type chamber which is 5 cm from the bottom of the chamber insert second to the certificate. The recombination and voltage correction factors were not applied to the measurement calculations.

The form we employ for the full calibration is on an Excel spreadsheet (Fig.2) which allows convenient calculations of source activity and positioning. The source is programmed to go to a series of positions within the well chamber and the maximum current reading is used to calculate the activity in air kerma units. Also the temperature and pressure are taking into account. The well-type chamber is placed into

\[
\mu U = 1 \mu\text{Gy} m^2 h^{-1}
\]
the treatment room generally 24 hours before measurement in order to have the same temperature as the room. This value is then compared to the manufacturer’s stated activity decayed to the day of measurement. Agreement is typically within 2%. The equipment calibrations traceable to a national standard are not available, but dosimetry system constancy checks can be performed using our ⁶⁰Co teletherapy units. This application was however only done two times. The accurate position and the physical dimensions of the source are done by visual test using a special ruler supplied by the manufacturer.

![Fig. 2 – calibration spreadsheet ¹⁹²Ir](image)

3. RESULTS

As shown in Fig.3 the Air Kerrma Rate measured with the Well-type ionization chamber HDR1000 Plus was within an agreement of 2% with the Air Kerma rate delivered by the source vendor.

![Fig. 3 – deviation of Air Kerma Rate measured vs provided](image)

Only seven sources were compared as we miss the dosimetry data of the source no1. The source no3 was not measured and not used at all because the machine was not working in that time. Eight sources were accepted for clinical use as the measurements were within the tolerance. One source of nine imported (source no3, not included in the chart) was not measured and not used as the machine was not working. The source no1 was within the tolerance but we are missing the measured values to compare with the others, so it is not included in the chart. The source no4 with a deviation of -1.92% has been measured with another electrometer due to which we think we have this difference compared to other sources measurements. We also compared the well chamber result in this case with a free in-air farmer type chamber measurement that gave us a deviation to source certificate of 4% that was still inside the tolerance to accept a source for clinical use. The deviations of measured Air Kerma rate to the value of the sources certificates of all our used ¹⁹²Ir sources are less than 2%, which are within the tolerance (Fig.4).

![Fig 4. RAKR measured vs provided corrected for time decay.](image)

4. CONCLUSION

The deviations of measured Air Kerma rate to the value of the sources certificates of all our used ¹⁹²Ir sources are less than 2%, which are within the tolerance. The checked value of updated source strength in planning system and afterloading units for all sources was done and was correctly calculated. A verification with other chambers has been done and also has given very good results. An IAEA external audit has acknowledged the good experience.

5. REFERENCES

[4] SEFM, Sociedad Española de Fisica Medica, Calibración, matrices de dosis y control de


IMPLEMENTATION OF RAY SAFE i2 SYSTEM FOR STAFF DOSE MEASURING IN INTERVENTIONAL RADIOLOGY

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Abstract – Interventional radiology procedures usually delivered the highest radiation dose to the patients as well as to medical personal. Beside another factors like patient size, fluoroscopy time, machine calibration etc., a good clinical practice has strong effects to staff and patient’s radiation dose.

Materials and methods: In August 2012, a Ray Safe i2 system was installed in a private hospital in Skopje. The main purpose of this dosimetry system is to provide real time indication for the current exposure level of the medical personal. Knowing that, the staff has prerequisites to adjust their behavior to minimize unnecessary exposure like changing distance from exposed volume, C-rang angulations, field of view etc. and on this way to develop a good clinical practice. The Ray Safe i2 system is consisted by ten digital dosimeters, two dock stations, real time display, dose viewer and dose manager software. During interventional procedures, each involved staff wears dosimeter which measures and records X-Ray exposure every second and transfer the data wirelessly to the real time display. Color indication bars (green, yellow, red) represents the intensity of the currently received exposure, whereas green zone indicates < 0.2 mSv/h, yellow zone from 0.2 to 2 mSv/h and red zone indications from 2 to 20 mSv/h. Additionally, accumulated dose per individual is displayed next to the color indication bars. By using the software, information about personal dose history, such as annual dose, dose per particular session, hour, day or week, can be viewed and analyzed.

Results: In this work it was found that staff accumulated doses were constantly increased over time, but reported number of procedures does not correspond to this tendency. Our assumption is that there is a misleading between reported number and actual performed procedures.

Doctor1 received 55 times more dose than Doctor2 and Nurse1 received 11 to 3 times more dose than another Nurses.

It was found a correlation of $R^2 = 0.1247$ and $p=1.71303E-12$ between DAP and fluoroscopy time regardless of type of procedure.

Conclusions: Staff doses are proportional to number of procedures and fluoroscopy time, but depend also on patient size, exposure factors, type of procedure, etc. The main reason for discrepancies among personal doses is based on different number of performed procedures. According to Ray Safe dose records, it is very unlikely that personal will reach the annual dose limits.

Keywords – Ray Safe i2 system, interventional radiology staff doses

1. INTRODUCTION

Interventional radiology procedures usually delivered the highest radiation dose to the patients as well as to medical personal. Beside another factors like patient size, fluoroscopy time, machine calibration etc., a good clinical practice has strong effects to staff and patient’s radiation dose. The RaySafe i2 System is an electronic X-ray dose monitoring system. The intended use is to improve the awareness of people who work with or are in the presence of X-ray imaging equipment, about their occupational dose (also known as staff dose). The awareness focuses on:

- A graphical visualization of the real-time staff dose rate while working with X-ray equipment in examination rooms during medical procedures;
- Instant access to historical staff dose for reporting and analysis purposes.

The benefits of the RaySafe i2 System are to:

- Make people aware of their received staff dose during clinical work with X-ray imaging equipment;
- Instantly visualize the result of reducing measures of occupational dose by, for example, changing a person’s position in the examination room.

2. MATERIALS AND METHODS

The RaySafe i2 System can contain the following components: dosimeter, real time display, dose viewer (computer software), dose manager (computer software) and cradle (dock station used to connect dosimeters and computer) [1].

Accumulated dose in relation to the annual dose limit for the current year. The annual dose limit for the dosimeter is dependent of the shielding factor of the lead apron, as well as other radiation protection used. The more protection the higher the limit can be for the same effective dose to the user. The annual dose limit may or may not be chosen to reflect the legal dose limit where it is used. It could reflect the legal limit or for example a lower target for the clinical user case. Assuming that lead upon accumulated 80% of radiation, we set an annual limit of 100 mSv.

More detailed historical dose information can be transferred from dosimeters via the cradle connected to a computer and viewed using the computer software (dose viewer and dose manager). The dose viewer software is also used for administrating dosimeters, change dosimeter names, colors and reset dose history. The dose manager software is an advanced software for analyzing, reporting and archiving dose information.

In August 2012, a Ray Safe i2 system was installed in private hospital in Skopje. The system was used by staff which performs cardiology procedures by using an X-ray angiography machine GE Innova 2100. Dose parameters data, data about the type and number of procedures, as well as fluoroscopic time and Dose Area Product (DAP) values, were collected for eight months, until March 2013.
3. RESULTS

Since September 2012 to March 2013, according to the collected data, it was performed total of 276 interventional cardiology procedures. Number of procedures by type is presented on Figure 4.

Diagnostic coronary angiography was the most often interventional cardiology procedure, while stenting was the most fluoroscopic time demanding procedure.

Staff dose rate and staff accumulated dose history since September 2012 through March 2013 are presented on the Figure 5-10 for each staff member. It is obvious that staff accumulated doses were constantly increased over time, but reported number of procedures does not correspond to this tendency. Our assumption is that there is a misleading between reported number and actual performed procedures.

In Table 1 are presented data about annual dose, percentage of set yearly maximum values and total received dose.

<table>
<thead>
<tr>
<th>Display name</th>
<th>Annual dose (mSv)</th>
<th>Yearly max (100 mSv)</th>
<th>Total dose (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Doctor1</td>
<td>30.9</td>
<td>30%</td>
<td>34.6</td>
</tr>
<tr>
<td>Doctor2</td>
<td>0.565</td>
<td>0%</td>
<td>1.76</td>
</tr>
<tr>
<td>Nurse1</td>
<td>23.1</td>
<td>23%</td>
<td>26.9</td>
</tr>
<tr>
<td>Nurse2</td>
<td>2.35</td>
<td>2%</td>
<td>6.6</td>
</tr>
<tr>
<td>Nurse3</td>
<td>7.5</td>
<td>7%</td>
<td>10.4</td>
</tr>
<tr>
<td>Nurse4</td>
<td>2.0</td>
<td>1%</td>
<td>5.01</td>
</tr>
</tbody>
</table>

Annual dose is received dose in 2013, while total dose is dose received from the beginning of dose measuring by the Ray Safe system.

On Figure 11 are presented DAP values related to fluoroscopic time for different type of procedures. It was found a correlating factor of $R^2 = 0.1247$ what indicates that identical fluoroscopy time for two different patients does not lead necessary to equal...
DAP values. Due to different patient size, projection views, field sizes etc., the machine dynamically adjust technical parameters like kV, mA etc. It means that patient or staff in some cases can receive much higher dose for shorter that for longer fluoroscopy time in another case. From our data in four cases there are big deviation in DAP values (indicates by red spots on the graph) whereas in two cases DAP values are very high and in the another two, very low, compare to trend values. A further investigation for these cases is needed.

![Graph showing correlation between DAP and fluoroscopy time]

*Fig. 11 – Correlation between DAP and fluoroscopy time*

In terms to assess the significance of correlation between DAP values and fluoroscopy time, a T.TEST was performed. It was obtain a value of \( p=1.71303 \times 10^{-12} \) what indicates that there is a statistically significance between both parameters.

4. CONCLUSION

Ray Safe i2 is a dosimetry system for measuring and displaying of dose rate and accumulated dose to personal during interventional procedure. Knowing of real time dose rate of each staff member can be useful to optimize clinical practice regards to radiation protection of staff and patients. Staff doses are proportional to number of procedures, fluoroscopy time, but also depend on patient size, exposure parameters, type of procedures etc.

5. REFERENCES

Unfurls Ray Safe 5001047-A

ACKNOWLEDGMENT

Authors express acknowledgment to the staff at Interventional Cardiology department in the private clinical hospital for using the Ray Safe i2 system in their clinical practice and for providing data about number and type of procedures.
SURVEY ON THE FREQUENCY OF TYPICAL X-RAY EXAMINATIONS AND ESTIMATION OF ASSOCIATED POPULATION DOSES IN THE REPUBLIC OF MACEDONIA

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Abstract – Purpose: Medical X-ray exposures have been the largest man-made source of population exposure to ionizing radiation in developed countries for many years. It is therefore important for radiation protection and healthcare authorities in each country to regularly assess the magnitude and the distribution of this large and increasing source of population exposure. The purpose of this paper is to present results from the survey on the number of typical X-ray examination procedures in the Republic of Macedonia for 2010, the distribution of examination procedures by type per modality for adults and pediatric patients, the annual frequency per 1000 population and the collective effective dose per 1000 population from the X-ray examination procedures performed in the Republic of Macedonia in 2010.

Materials and methods: In the beginning of 2011, a survey was initiated in the Republic of Macedonia for collecting data on the number of typical X-ray examination procedures conducted in 2010 as a basis for estimating frequency of these procedures and associated population doses. The survey was initiated within a Dose Data Med project¹ launched by the European Commission to study population doses from medical exposures within the Union. The Republic of Macedonia was invited to participate in this project as a test country². Typical X-ray examination procedures encompass those that are recognized to be the most important for the total population dose, referred to as TOP20 X-ray procedures. The survey was based on a specific questionnaire being prepared and distributed to the 87 X-ray departments in the Republic of Macedonia intended to cover the data for the year of 2010. The data was collected and summarized. Based on data gathered, the total number of examination procedures, annual frequency and their distribution by modality were calculated. Thereafter, the annual collective effective dose per 1000 population for each examination procedure in the TOP20 group and collective effective doses were estimated using literature data for values of the mean effective dose per typical examination procedure. Finally, normalization of the total collective effective dose from all TOP 20 X-ray procedures for the whole population in the Republic of Macedonia was performed.

Results: 67% of X-ray departments present in the Republic of Macedonia at the time the survey was initiated provided data on the number of TOP20 X-ray examination procedures performed in 2010. On the basis of the data gathered, a total of 322039 TOP20 X-ray examination procedures were performed in 2010 for both adult and pediatric patients. Plain radiography examination procedures (dental excluded) were the most commonly performed procedures in the Republic of Macedonia that year and the plain radiography of chest/thorax had the highest frequency of examinations (64 examinations) per 1000 population. The Ba meal examination procedure with an annual frequency of 2.93 per 1000 population has the highest contribution to the annual collective effective dose of all other procedures. Still, in total, the contribution of X-ray examinations in the plain radiography modality to the collective effective dose is the highest. The total collective dose from TOP 20 X-ray examination procedures in 2010 is 507 man Sv, while the normalized collective dose to the population is 249.7 mSv/1000 population.

¹ So-called DDM2 project
² CLAIM: No financial compensation, travel grant or any other benefit was provided for participants from the Republic of Macedonia in this project.
Conclusions: The most common type of examination in the Republic of Macedonia for 2010 is X-ray projection of lungs. The contribution to the collective effective dose from X-ray examinations in the plain radiography modality is the highest, followed by contributions from fluoroscopy procedures, computer tomography and interventional radiology procedures. Comparison of the estimated collective dose from TOP20 X-ray examination procedures in other countries suggests possible underestimation in the estimated doses comparing to actual doses. A more comprehensive survey and analysis are needed to be carried out in future in order to obtain more precise estimates of the frequency of X-ray examinations and associated population doses in the Republic of Macedonia. Ideally, this need to be done on a regular basis with a commitment by relevant institutions.

Keywords – frequency of X-ray examination, population dose

1. INTRODUCTION

Medical X-ray exposures have been the largest man-made source of population exposure to ionizing radiation in developed countries for many years. Recent developments in medical imaging, particularly with respect to computed tomography (CT), have led to rapid increases in the number of relatively high-dose X-ray examinations performed, with significant consequences for individual patient doses and for the collective dose to the population as a whole. It is therefore important for radiation protection and healthcare authorities in each country to regularly assess the magnitude and distribution of this large and increasing source of population exposure.

In the beginning of 2011, the European Commission launched a “Study on European Population Doses from Medical Exposures” so call DDM2 project. The predominant objectives of these population dose assessments in recent years have been:

1. To observe trends in the annual collective dose and the annual average per caput dose from medical x-rays in a country with time (per caput dose = collective dose averaged over the entire population);

2. To determine the contributions of different imaging modalities and types of examination to the total collective dose from all medical x-rays;

3. To determine the relationship between the frequencies of different types of x-ray examination, the typical radiation doses given to patients and their contribution to the total collective dose;

4. To determine whether there are any regional variations within a country in the frequency and per caput dose from particular types of x-ray examination and

5. To compare the frequencies and the annual per caput doses from medical x-rays between countries.

Republic of Macedonia has been invited to participate in the project as a test country among other five test countries.

2. MATERIALS AND METHODS

Macedonia has participated in DDM2 project in X-ray and Nuclear Medicine procedures. This paper presents results for X-ray procedures only. The first step was answering of a general questionnaire about national regulatory framework and assessment of the status of implementation of the requirements for medical dose surveys and population dose estimations. According to the available categorization of X-ray examination procedures, it was decided that Macedonia will collect for data on the TOP20 X-ray examinations presented in Table 1. Additionally are presented average mean effective doses what were estimated in the first Dose Data Med project (DDM1).

<p>| Table 1. TOP20 X-ray examination and mean effective dose per examination DDM1 project [1] |</p>
<table>
<thead>
<tr>
<th>Category (modality)</th>
<th>Exam type</th>
<th>Mean eff.doses (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Plain radiography</td>
<td>1. Chest / thorax</td>
<td>0.25</td>
</tr>
<tr>
<td></td>
<td>2. Cervical spine</td>
<td>0.7</td>
</tr>
<tr>
<td></td>
<td>3. Thoracic spine</td>
<td>2.0</td>
</tr>
<tr>
<td></td>
<td>4. Lumbar spine</td>
<td>2.8</td>
</tr>
<tr>
<td></td>
<td>5. Mammography</td>
<td>0.4</td>
</tr>
<tr>
<td></td>
<td>6. Abdomen</td>
<td>1.8</td>
</tr>
<tr>
<td></td>
<td>7. Pelvis &amp; hip</td>
<td>1.35</td>
</tr>
<tr>
<td>Radiography / Fluoroscopy</td>
<td>8. Ba - meal</td>
<td>15.0</td>
</tr>
<tr>
<td></td>
<td>9. Ba - enema</td>
<td>12.5</td>
</tr>
<tr>
<td></td>
<td>10. Ba – follow</td>
<td>24.5</td>
</tr>
<tr>
<td></td>
<td>11. IVU</td>
<td>3.5</td>
</tr>
<tr>
<td></td>
<td>12. Cardiac angiography</td>
<td>11.25</td>
</tr>
<tr>
<td>Computed Tomography (CT)</td>
<td>13. CT head</td>
<td>2.4</td>
</tr>
<tr>
<td></td>
<td>14. CT neck</td>
<td>2.8</td>
</tr>
<tr>
<td></td>
<td>15. CT chest</td>
<td>8.2</td>
</tr>
<tr>
<td></td>
<td>16. CT spine</td>
<td>6.0</td>
</tr>
<tr>
<td></td>
<td>17. CT abdomen</td>
<td>13.5</td>
</tr>
<tr>
<td></td>
<td>18. CT pelvis</td>
<td>8.8</td>
</tr>
<tr>
<td></td>
<td>19. CT trunk</td>
<td>24.4</td>
</tr>
<tr>
<td>Interventional radiology</td>
<td>20. PTCA</td>
<td>15.35</td>
</tr>
</tbody>
</table>

The second step was preparing a national questionnaire for reporting the number of TOP20 X-ray examinations and guideline in support to the questionnaire that describes more accurately what and how needs to be reported. The questionnaire
including the guideline was sent to all 87 X-ray departments in the Republic of Macedonia. The data reported was collected and the total number of examinations summarized. Thereafter, the annual frequency of X-ray examinations per year was calculated and the distribution by modality was determined by:

\[
Annual\ frequency = \frac{\text{Total number of examinations}}{\text{population}} \times 1000 \tag{1}
\]

Using the annual frequency of X-ray examinations and multiplying by mean effective dose per examinations stated in reference [1], the annual collective effective dose per 1000 population for each examination type in TOP20 group was calculated by:

\[
\text{Annual coll eff dose} = \text{Total number} \times \frac{\text{mean eff dose}}{\text{per exam}} \tag{2}
\]

Finally, contribution to the collective effective dose from TOP20 group by modality was calculated.

Earlier UNSCEAR reports expressed patient doses in terms of the mean absorbed dose to few organs or tissues (e.g. the gonads and red bone marrow) and population doses were expressed in terms of the annual genetically significant dose and the annual per caput red bone marrow dose. In its more recent reports [UNSCEAR, 1993 and 2000] UNSCEAR has used effective dose [ICRP, 1991] as a convenient indicator of overall risk-related exposure of the patient from an X-ray examination, and population doses were expressed in terms of the annual collective effective dose per 1000 population. The effective dose (E) essentially takes account of non-uniform body exposures and organs and tissues which are sensitive to radiation by estimating the average whole body dose that would result in the same total radiation-induced cancer risk as the non-uniform body exposure. The collective effective dose (S) takes account of the number of people exposed to a particular source by multiplying the average effective dose to the exposed group by the number of individuals in the group. Since the collective population dose depends on the size of the population, it is often more useful to use the annual average per caput dose (i.e. the annual collective dose averaged over the entire population). However, for the purpose of making population dose estimates, it is reasonable to assume that children receive the same mean effective dose as adults from the same type of examination. When exposure factors are selected to suit the smaller sizes of pediatric patients and to maintain the same dose to the image receptor, entrance surface doses will be smaller than for adults but will be attenuated less to reach organs at depth, resulting in similar effective doses. For this reason, patient dose surveys was designed to determine mean effective doses for use in population dose estimates taking into account adult patients only. [1]

It is recognized that certain limitations and uncertainties in the estimated frequency of TOP20 examinations and population doses exists due to:

- Percentage of reported number of examinations;
- Insufficiently differentiated codes of examinations;
- Lack of data from some important providers of radiology;
- Mistakes occurring in collected or recorded data;
- Use of literature data for the mean effective dose per examination;
- Uncertainties in conversion coefficients, etc.

For example, considering the fact that only 67% of all X-ray departments present in the Republic of Macedonia at the time the survey was initiated responded to the survey, the uncertainties of reported results are ±25%. Therefore, more comprehensive survey and analysis needs to be carried out in future in order to obtain more precise estimates of the frequency of X-ray examinations and associated population doses.

3. RESULTS

Fifty-seven X-ray departments of total of 85 departments (67%) present at the time the survey was initiated provided data on the number of TOP20 X-ray examination procedures. In these departments, a total of 322039 TOP20 X-ray examination procedures were performed in 2010. 91.3% of patients engaged in these examinations were adults and 8.69% pediatric patients younger than 14 year.

In this paper, the following results are presented: TOP20 X-ray examinations distribution by modalities, the distribution of examinations in radiography, fluoroscopy, interventional radiology and Computed Tomography (CT) by types; the estimated annual frequency and collective dose per 1000 population; and the estimated collective dose from TOP 20 X-ray examinations.

An overview of the number and type of machines for medical imaging present in the Republic of Macedonia is presented on Figure 1.

![Image](image1.png)

\( \text{Fig. 1 – Number of X-ray and other type of machines for medical imaging present in the Republic of Macedonia} \)
3.1. Examinations by modality

Plain radiography (dental excluded) examination procedures are the most commonly performed procedures in the Republic of Macedonia (Figure 1).

![Fig. 2 – X-Ray examinations distribution by modality](image)

The total number of TOP20 X-ray examinations per modality was summarized considering adult patients only. On the basis on this number, the annual frequency per 1000 population was calculated. The results are presented in Table 2.

<table>
<thead>
<tr>
<th>Modality</th>
<th>Total number</th>
<th>Annual frequency per 1000 popul.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Plain radiography (dental excluded)</td>
<td>257 482</td>
<td>126.84</td>
</tr>
<tr>
<td>Radiography / Fluoroscopy</td>
<td>9 860</td>
<td>4.86</td>
</tr>
<tr>
<td>Cardiac Angiography</td>
<td>3 029</td>
<td>1.49</td>
</tr>
<tr>
<td>Computed Tomography</td>
<td>16 856</td>
<td>8.30</td>
</tr>
<tr>
<td>PTCA</td>
<td>900</td>
<td>0.44</td>
</tr>
</tbody>
</table>

According to the collected data, 288 127 TOP20 X-ray examinations were performed on adult population in 2010 in the Republic of Macedonia.

3.2. Examinations by type in different modalities

3.2.1. Radiography modality

The most common type of examination is X-ray projection of lungs (26.35% of all examinations in adults, Figure 3, and 40.72% of all examinations in pediatric patients, Figure 4). In Computed Tomography (CT), the most frequent examination is head examination (53% of all examinations in adults and 48% of all examinations in pediatric patients).

![Fig. 3 - Distribution of radiography examinations in adult patients](image)

![Fig. 4 - Distribution of radiography examinations in pediatric patients (<14 y)](image)

3.2.2. Fluoroscopy

An examination of the stomach and the duodenum with Ba meal is the most common type of fluoroscopy examination among both adult and pediatric patients.

![Fig. 5 - Distribution of fluoroscopic examinations in adult patients](image)

![Fig. 6 - Distribution of fluoroscopic examinations in pediatric patients (< 14 y)](image)
3.2.3. Interventional radiology

The diagnostic coronary angiography is the most common procedure in interventional radiology.

![Fig. 7 - Distribution of Interventional radiology examinations in adults patients](image)

3.2.4. Computed Tomography

CT head scans in both adult and pediatric patients are the most common CT procedures.

![Fig. 8 - Distribution of CT examinations in adult patients](image)

![Fig. 9 - Distribution of CT examinations in pediatric patients (<14 y)](image)

3.3. Annual frequency per 1000 population

It was made an interpolation of results as all 87 X-ray units participated in the survey and the annual frequency per 1000 population was calculated for adult patients only (Figure 10). Interpolated results show that 64 exams of chest or thorax, 24.89 for pelvis and hip and 12.30 for lumbar spine (including limbo-sacral joint) were performed per 1000 population in 2010.

![Fig. 10 - Frequency per 1000 population for TOP20 X-ray examinations in 2010](image)

Plain radiography of chest/thorax had the highest frequency of examination per 1000 population in 2010 in the country.

3.4. Annual collective effective dose

The annual collective effective dose per examinations in TOP20 group were calculated for 2010 (Figure 11). The Ba meal examination procedure with an annual frequency of 2.93 per 1000 population, had the highest contribution to the annual collective effective dose with 89.22 manSv. Among highest contribution to the collective dose, have lumbar spine examination with 69.90 manSv and pelvis & hip examination with 68.22 manSv, while CT scans of abdominal regia have a 28.63 manSv.

![Fig. 11 - Annual collective effective dose per type of X-ray examinations in TOP20 group](image)

Due to the high frequency of Ba meal examination procedure, this examination had the highest contribution to the collective dose per examination. Radiography of lumbar spine as well as pelvis and hip, had also a big contribution to the collective dose per examination in 2010.
In total, the contribution of X-ray examinations in plain radiography modality to the collective effective dose for 2010 is the highest. Although the patient dose during interventional radiology procedures are much higher comparing to those from plain radiography procedures, in total the examinations in interventional radiology have no the highest contribution to the collective effective dose. Comparing results with another countries, the contribution from CT scans in our country is lower, probably due to less reported number of performed exams.

4. CONCLUSION

In the Republic of Macedonia, there is no system in place for gathering data on national level for the number and type of X-ray examinations procedures performed and for estimating the associated population doses. This survey and analysis carried out within the DDM2 project is the very first intent in the country for doing so. Considering the limitations and uncertainties associated with the collected data and estimations made, a more comprehensive survey and analysis needs to be carried out in future (ideally, on a regular basis) in order to obtain more precise estimates of the frequency of X-ray examinations and associated population doses.

The most common type of examination in the Republic of Macedonia is the X-ray projection of lungs/thorax. Macedonian population received the highest dose from the examinations in plain radiography modality due to their high frequency for 2010.

The frequency of TOP20 X-ray examinations per 1000 population in the Republic of Macedonia is higher only than Romania and Cyprus [1]. This implies a possibility that the number of reported X-ray examinations in TOP20 group were lower than the actual. As a consequence, there is a possibility for the calculated doses to be lower than the actual doses.

5. REFERENCES


ACKNOWLEDGMENT

Authors express their gratitude to the 57 X-ray departments in the Republic of Macedonia that took part in the survey. Without their participation in the survey, this analysis would not have been possible.
FLUENCE COMPLEXITY FOR IMRT FIELD AND SIMPLIFICATION OF IMRT VERIFICATION

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² Proton Therapy Center Czech, Budínova 2437/1a, 180 00 Praha 8, Czech Republic, vladimir.vondracek@ptc.cz
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Abstract – Intensity Modulated Radiation Therapy (IMRT) requires dosimetric verification of each patient’s plan, which is time consuming. This work deals with the idea of minimizing the number of fields for control, or even replacing plan verification by machine quality assurance (QA). We propose methods for estimation of fluence complexity in an IMRT field based on dose gradients and investigate the relation between results of gamma analysis and this quantity. If there is a relation, it might be possible to only verify the most complex field of a plan. We determine the average fluence complexity in clinical fields and design a test fluence corresponding to this amount of complexity which might be used in daily QA and potentially replace patient-related verification. Its applicability is assessed in clinical practice. The relation between fluence complexity and results of gamma analysis has been confirmed for plans but not for single fields. There is an agreement between the suggested test fluence and clinical fields in the average gamma parameter. A critical value of average gamma has been specified for the test fluence as a criterion for distinguishing between poorly and well deliverable plans. It will not be possible to only verify the most complex field of a plan but verification of individual plans could be replaced by a morning check of the suggested test fluence, together with a well-established set of QA tests.

Keywords – IMRT verification, gamma analysis, fluence complexity, plan deliverability

1. INTRODUCTION

The radiotherapeutic technique IMRT (Intensity Modulated Radiation Therapy) allows to increase dose to tumor and decrease radiation damage to organs at risk (OARs). Thus, it allows better tumor control. However, due to the increased dose, there is a higher risk of damage to OARs in case of wrong dose delivery and the whole process requires precise verification. This is time-consuming and often an argument against the use of IMRT at busy clinics. The question is whether it is possible to simplify the verification process without increasing the risk for patient.

There might be a possibility to reduce the time needed for patient-related verification. Let us suppose that the results of verification of a particular field correspond to the fluence complexity of the field. Then it would be enough to verify only one field of the patient’s plan – the one with the most complex fluences. If this field met the tolerance criteria, we could say that other fields meet them automatically. There is yet another solution: to determine the average fluence complexity of a clinical IMRT field and create a test fluence that would be close to this amount of complexity and that would be verified during the morning checks of the linear accelerator. This test, together with extended quality assurance (QA) of the multileaf collimator (MLC), could potentially replace patient-related verification.

Several means of fluence complexity definition have been published, among them the Modulation Index (MI) or the Modulation Complexity Score (MCS). However, different research groups publish different, conflicting results because their IMRT verification process differs. Thus, it is difficult to find a universal definition of fluence complexity which would be suitable to evaluate dose delivery at all clinics.

This work defines fluence complexity in a new way. We look for a correlation between thus defined quantity and results of gamma analysis of IMRT fields. We try to confirm the hypothesis that the more complex the fluences are, the worse results of gamma
analysis we get. Based on a set of clinical plans, we calculate the average fluence complexity in an IMRT field and we design a test fluence with similar complexity which can be used for morning checks of the linear accelerator. Suitability of this fluence is tested in clinical practice.

2. MATERIALS AND METHODS

Works that have been published so far calculate fluence complexity in an IMRT field using plan parameters such as the optimal fluence estimated by the treatment planning system (TPS) or the number of monitor units (MU). The most frequently used parameters are probably the Modulation Index [1, 2] and the Modulation Complexity Score [3]. Recently, another promising solution has been proposed by Nauta et al. [4] using fractal analysis. Here we define fluence complexity using the number and the amplitude of dose gradients in matrices of dose distribution calculated by the Portal Image Prediction algorithm (PDIP) [5] in the plane of the Electronic Portal Imaging Device (EPID) when creating a verification plan [6]. Mathematically, the quantity can be defined as

\[ P_q = \sum_{i=1}^{m} \sum_{j=1}^{n} k_{ij}, \quad k_{ij} = \begin{cases} 1 & c_{ij} > q \\ 0 & c_{ij} \leq q \end{cases} \]

\[ V_q = \sum_{i=1}^{m} \sum_{j=1}^{n} k_{ij}, \quad k_{ij} = \begin{cases} c_{ij} & c_{ij} > q \\ 0 & c_{ij} \leq q \end{cases} \]

(1)

where \( P_q \) is the number of dose gradients in the matrix mentioned above which are greater than a certain limit \( q \), \( m \) and \( n \) are the matrix dimensions\(^1\) and \( c_{ij} \) is the amplitude of the dose gradient on the position \([i, j]\). The latter is calculated as the difference between values in adjacent pixels divided by the real distance between measuring points of the detector (Fig. 1). Thus, for each measuring point we obtain 8 values of dose gradient (one in each of the 8 directions). The value of \( q \) was chosen to be 0, 200, 300 and 400 arbitrary units.\(^2\) (We suppose that small gradients are present everywhere and equally distributed, so they do not affect fluence complexity in a significant way.)

Table 1. Gamma analysis tolerance limits

<table>
<thead>
<tr>
<th>Dose difference</th>
<th>DTA</th>
<th>Maximum gamma</th>
<th>Average gamma</th>
<th>Area gamma</th>
</tr>
</thead>
<tbody>
<tr>
<td>3 [%]</td>
<td>3 mm</td>
<td>( \leq 3.5 )</td>
<td>( \leq 0.3 )</td>
<td>3%</td>
</tr>
</tbody>
</table>

\(^1\) The expression \( \delta n \) occurs in the formula due to the way the calculation algorithm is implemented in MATLAB.

\(^2\) Values in the predicted matrix of dose distribution which is then exported in DICOM format from the TPS Eclipse are relatively proportional to dose. However, no physical quantity can be assigned to them.

Matrices of dose distribution predicted in the plane of the detector EPID were exported in DICOM format from the TPS Eclipse version 8.6 (Varian Medical Systems, Palo Alto, USA). MATLAB was used to calculate fluence complexity. The experiment was carried out with the EPID aS500 and the Varian CLINAC 600C/D 6X (Varian Medical Systems, Palo Alto, USA).

First the correlation between fluence complexity in an IMRT field and results of gamma analysis [7, 8, 9] was observed. A statistical set of 100 IMRT plans for head-and-neck patients, that is 1310 fields, was used. Tolerance levels for gamma analysis at our institution are listed in Table 1, the dose difference criterion being the percentage of near-maximum planned dose [6]. The Spearman rank correlation coefficient was calculated to observe whether the rank of fields in a plan according to their fluence complexity is the same as the rank of the fields according to the gamma parameters. The percentage of plans where the correlation was significant was determined.

The average fluence complexity of a clinical IMRT field was estimated from the same statistical set of plans. It was determined by the total number of dose gradients in a field and by the average amplitude of one gradient in a field. Two test fluences that correspond to this average value of complexity were designed using the software Shaper (Varian Medical Systems, Palo Alto, USA). By a test fluence we mean a field of the size 10 x 10 cm which has a simple geometrical shape. These test fluences were verified in clinical practice over a one month period. With each patient plan verification, these fluences were delivered as well and the correlation between the gamma parameters of clinical plans and these test fluences was determined, again using the rank correlation coefficient.

3. RESULTS

For each of the 100 clinical IMRT plans the rank correlation coefficient was determined. The percentage of plans for which the correlation between the gamma parameters and fluence complexity in a field was significant was estimated. Table 2 and Table 3 show the results of correlation analysis for the case \( q = 0 \) and for the significance level \( \alpha = 0.05 \).
Table 2. Results of correlation analysis for IMRT fields

<table>
<thead>
<tr>
<th>Type of correlation:</th>
<th>Percentage of correlated data [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\gamma_{max}$ / Gradient amplitude</td>
<td>2</td>
</tr>
<tr>
<td>$\gamma_{max}$ / Number of gradients</td>
<td>3</td>
</tr>
<tr>
<td>$\gamma_{ave}$ / Gradient amplitude</td>
<td>25</td>
</tr>
<tr>
<td>$\gamma_{ave}$ / Number of gradients</td>
<td>37</td>
</tr>
<tr>
<td>$\gamma_{area}$ / Gradient amplitude</td>
<td>16</td>
</tr>
<tr>
<td>$\gamma_{area}$ / Number of gradients</td>
<td>12</td>
</tr>
</tbody>
</table>

Table 3. Results of correlation analysis for IMRT fields

<table>
<thead>
<tr>
<th>Type of correlation:</th>
<th>Percentage of correlated data [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\gamma_{max}$ / Gradient amplitude</td>
<td>1</td>
</tr>
<tr>
<td>$\gamma_{max}$ / Number of gradients</td>
<td>2</td>
</tr>
<tr>
<td>$\gamma_{ave}$ / Gradient amplitude</td>
<td>18</td>
</tr>
<tr>
<td>$\gamma_{ave}$ / Number of gradients</td>
<td>17</td>
</tr>
<tr>
<td>$\gamma_{area}$ / Gradient amplitude</td>
<td>9</td>
</tr>
<tr>
<td>$\gamma_{area}$ / Number of gradients</td>
<td>5</td>
</tr>
</tbody>
</table>

In the next step, the average fluence complexity in clinical IMRT fields was estimated by the average value of one dose gradient and the average number of gradients in a field (Table 4).

Table 4. Average fluence complexity in a clinical IMRT field

<table>
<thead>
<tr>
<th>$q$</th>
<th>Average gradient amplitude [a.u./mm]</th>
<th>Variance [a.u.$^2$/mm$^2$]</th>
<th>Standard deviation [a.u./mm]</th>
<th>Relative standard deviation [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>29</td>
<td>18</td>
<td>4</td>
<td>15</td>
</tr>
<tr>
<td>200</td>
<td>304</td>
<td>394</td>
<td>20</td>
<td>7</td>
</tr>
<tr>
<td>300</td>
<td>409</td>
<td>673</td>
<td>26</td>
<td>6</td>
</tr>
<tr>
<td>400</td>
<td>509</td>
<td>1020</td>
<td>32</td>
<td>6</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>$q$</th>
<th>Average number of gradients</th>
<th>Variance</th>
<th>Standard deviation</th>
<th>Relative standard deviation [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>74619</td>
<td>4 x 10$^8$</td>
<td>20449</td>
<td>27</td>
</tr>
<tr>
<td>200</td>
<td>2371</td>
<td>590528</td>
<td>768</td>
<td>32</td>
</tr>
<tr>
<td>300</td>
<td>889</td>
<td>136846</td>
<td>370</td>
<td>42</td>
</tr>
<tr>
<td>400</td>
<td>356</td>
<td>38878</td>
<td>197</td>
<td>55</td>
</tr>
</tbody>
</table>

Two test fluences were designed in the programme Shaper, so that they correspond to the average value of fluence complexity (Fig. 2). They were tested in clinical practice over a month. Table 5 summarizes the results of correlation analysis - they show the relation between the gamma parameters of the test fluences and of clinical IMRT fields. A significant correlation was observed for the parameter average gamma, even at the significance level 99 % (Fig. 3).

Table 5. Results of correlation analysis for the test fluences. Values of the Spearman correlation coefficient for the gamma parameters of clinical fields and the test fluences. Two measurements were performed for each test fluence every day. Significant correlation is highlighted

<table>
<thead>
<tr>
<th>Spearman correlation coefficient</th>
<th>test fluence no. 1</th>
<th>test fluence no. 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>maximum gamma</td>
<td>-0.1</td>
<td>-0.1679</td>
</tr>
<tr>
<td>average gamma</td>
<td>0.35</td>
<td>0.3536</td>
</tr>
<tr>
<td>area gamma</td>
<td>-0.0893</td>
<td>-0.075</td>
</tr>
<tr>
<td>Critical value</td>
<td>$\alpha = 0.05$</td>
<td>0.5179</td>
</tr>
</tbody>
</table>

Even though a significant correlation was confirmed between the test fluences and clinical IMRT fields in the parameter average gamma, the tolerance threshold of this parameter used clinically was never exceeded by the test fluences (while clinical fields did sometimes exceed the threshold). Therefore, ROC analysis (Receiver Operating Characteristic) was performed in order to estimate a lower tolerance threshold for average gamma of the test fluences. If the test fluence exceeded this threshold during the morning check of the linear accelerator, it would mean that clinical plans would not meet the tolerance criteria on that particular day, either. The best value was found to be $\gamma_{ave} = 0.210$ (Table 6). This test
revealed non-deliverable plans\(^3\) with 94% sensitivity but only 38% specificity\(^4\).  

The ROC curve is shown in Figure 4. The AUC parameter (Area Under Curve) has the value of 0.84, which means a very good test.

\[ \begin{array}{cccccc}
\text{Threshold} & \text{TP} & \text{FP} & \text{TN} & \text{FN} & \text{Sensitivity} & \text{Specificity} \\
\gamma_{\text{ave}} \geq 0.235 & 0 & 0 & 16 & 18 & 0.00 & 1.00 \\
\gamma_{\text{ave}} \geq 0.215 & 15 & 4 & 12 & 3 & 0.83 & 0.75 \\
\gamma_{\text{ave}} \geq 0.210 & 17 & 10 & 6 & 1 & 0.94 & 0.38 \\
\gamma_{\text{ave}} \geq 0.200 & 17 & 13 & 3 & 1 & 0.94 & 0.19 \\
\gamma_{\text{ave}} \geq 0.180 & 18 & 16 & 0 & 0 & 1.00 & 0.00 \\
\end{array} \]

\(^3\) A non-deliverable plan is a plan where the parameter average gamma exceeds the tolerance level at least in one of the fields.

\(^4\) The test correctly identified non-deliverable plans in 94% of cases but it only correctly identified deliverable plans in 38% of cases.

**Fig. 3 - Correlation of average gamma for clinical fields and the test fluences**

**Table 6. Sensitivity and specificity for different threshold values of \(\gamma_{\text{ave}}\) for the test fluence no. 2. TP – true positive, FP – false positive, TN – true negative, FN – false negative**

**4. CONCLUSION**

A new method for estimation of fluence complexity in an IMRT field has been proposed, which differs from those already published [1, 2, 3, 10]. It is based on matrices of dose distribution predicted for EPID in a verification plan. Fluence complexity is calculated as the amplitude and the number of dose gradients in an IMRT field.

In contrast to most papers, this publication uses a much larger set of data (100 plans, 1310 fields) and verifies the new method using clinical plans and clinically applied tolerance levels for gamma analysis. In a way, the method proposed here is similar to the MI parameter but it uses a different type of matrix.

A correlation between fluence complexity of IMRT fields and results of gamma analysis has been confirmed for a certain percentage of plans. However, this percentage is not big enough to only verify the most complex field of a plan, as intended.

For that reason, the average fluence complexity of clinical fields has been assessed and a test fluence of which the complexity is equal to the average value has been created. The test fluence has been verified in clinical practice and has been found to correlate well with clinical IMRT plans in the parameter average gamma. A critical value of average gamma has been proposed and if the test fluence exceeds it, it will mean that clinical plans will not meet the tolerance criteria, either, due to malfunction of the MLC or the EPID. This method has a high sensitivity in determining poorly deliverable plans, although due to low specificity it recognizes some plans as false positives.

The aim of this work was to simplify the process of IMRT verification. This can be done by replacing patient-related verification with a set of QA tests of
the MLC and the EPID, one of which would be a morning check of the proposed test fluence.

In future experiments it would be desirable to include other tumor localities, e.g. prostate and brain. The test fluence should be verified in clinical practice over a longer period of time. If the methods proposed here were to be used at all institutions, it would be necessary to include other detectors, because the EPID detector is not available at all clinics.

5. REFERENCES


LOCALLY-REGIONALLY ADVANCED TONSILLAR SQUAMOUS CELL CARCINOMA TREATED WITH CONCURRENT CHEMORADIOTHERAPY

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Abstract – Purpose: To perform a retrospective review of stage III–IV squamous cell carcinoma of the tonsil managed by definitive concurrent chemoradiotherapy (CCRT) in order to analyze the patients’ outcome and to evaluate the acute and late toxic effects of this treatment modality.

Material and methods: Between January 2005 and December 2010, 36 patients with locally and/or regionally advanced tonsillar cancer underwent three dimensional conformal radiotherapy (3DCRT) with concurrent platinum-based chemotherapy. The dose prescription of the planning target volume for gross tumor and low-risk subclinical disease was 70 Gy and 50 Gy, respectively. Conventional fractionation with a daily dose of 2.0 Gy, 5 times per week was used. Concurrent chemotherapy consisted of cisplatin 30 mg/m² given on a weekly basis. Acute and late radiotherapy-related toxicities were recorded using European Organization for Research and Treatment of Cancer/Radiation Therapy Oncology Group (EORTC/RTOG) grading system. The 3-year locoregional relapse-free survival (LRRFS), disease-free survival (DFS), and overall survival (OS) rates were calculated using the Kaplan-Meier method.

Results: The median follow-up of all patients was 20.5 months (range, 5 to 90 months). The median follow-up of living patients was 59 months (range, 30 to 90 months). Complete response rates of the primary tumor and of the nodal disease were 72.2% and 64.0%, respectively. A complete composite response was present in 25 patients (69.4%). Treatment failure occurred in 15 out of 25 patients who achieved complete composite response following CCRT. The 3-year LRRFS, DFS, and OS rate was 38.8%, 27.8%, and 27.3%, respectively. Grade 3 mucositis occurred in 58.3% of patients. Xerostomia grade 2 was revealed in 72.2% of patients.

Conclusion: Taking into account the low 3-year survival rates observed in our study and the high percentage of grade 2 xerostomia, it can be concluded that in the future, instead of 3DCRT with concurrent chemotherapy, intensity modulated radiation therapy (IMRT) in combination with concomitant chemotherapy should be considered a fully recommendable treatment option for advanced tonsillar carcinoma in order to improve locoregional control and survival with simultaneous reduction of late salivary toxicity and therefore prevention of late xerostomia.

Keywords – tonsillar carcinoma, radiotherapy, chemotherapy, concurrent chemoradiotherapy, xerostomia

1. INTRODUCTION

Squamous cell carcinoma of the tonsil represents the most common oropharyngeal cancer with the incidence described as steadily increasing due in part to human papilloma virus infection [1–4]. The tonsils are characterized by deep crypts in which squamous cell carcinomas may arise without causing obvious surface ulceration which enables small tumors to cause only little distress and remain unnoticed by the patient for a substantial period of time [5, 6]. The dominantly manifested advanced stage of patients’ disease at admission is a consequence of the anatomic location of tonsillar carcinoma, as well as of its origin in the lymphatic field of the oropharynx [7, 8]. There is a high frequency of nodal involvement at presentation of the patients with tonsillar cancer. Hence, Perez et al. [9] in their study reported that 58% of patients were initially diagnosed with metastases in the lymph nodes of the neck.

The treatment for advanced but resectable tonsillar carcinoma has traditionally been radical surgery and postoperative radiotherapy. Radiotherapy alone has
been the treatment approach used for advanced unresectable lesions and has been accepted also as an alternative to surgery in patients with resectable lesions in order to provide organ preservation. In recent years, based upon the results of several randomized studies and meta-analyses, chemotherapy administered concomitantly with radiotherapy has become the standard of care for advanced head and neck cancer [10-15]. The results of the French Head and Neck Oncology and Radiotherapy Group (GORTEC 94-01) phase III randomized trial favoring the use of chemoradiotherapy vs. radiotherapy alone for advanced stage oropharyngeal cancer [16, 17] have also confirmed that definitive CRT should be considered a recommended treatment option for advanced stage tonsillar squamous cell carcinoma.

As radiotherapy technique employed in the treatment of tonsillar carcinoma has evolved from two-dimensional radiotherapy (2DRT) to three-dimensional conformal radiotherapy (3DCRT), a significant improvement of tumor coverage and the capacity for sparing sensitive organs has been shown [18]. Intensity-modulated radiotherapy (IMRT) achieving higher total doses in tumors by delivering larger doses per fraction to the tumor only, has also been shown as an effective treatment technique for locally advanced oropharyngeal carcinoma [19], offering excellent results of locoregional control while limiting the dose to the surrounding critical tissues [20-23].

In order to summarize the results of non-surgical treatment approach we retrospectively analyzed patients with locally-regionally advanced tonsillar cancer treated with 3DCRT and concurrent chemotherapy.

2. MATERIALS AND METHODS

Thirty six patients with stage III or IV tonsillar cancer were treated with radiotherapy concurrent with platinum-based chemotherapy from January 2005 to December 2010 at the University Clinic of Radiotherapy and Oncology in Skopje. All patients had histologically proven squamous cell carcinoma of the tonsil and no radiologic evidence of distant metastases. Pretreatment evaluation other than medical history and clinical examination consisted of fiberoptic endoscopy with biopsy to obtain the histological proof, fine-needle aspiration biopsy in cases with detectable neck adenopathy, complete blood count, basic blood chemistry, chest x-ray, abdominal ultrasound and bone scan. The disease evaluation also included computed tomography (CT) scanning and/or magnetic resonance imaging (MRI) of head and neck region. Patients were staged according to the 2002 classification of the American Joint Committee on Cancer Staging (AJCC) [24].

2.1. Treatment

2.1.1. Radiotherapy

For 3DCRT, we used the Eclipse Version 7.3.10, a commercial 3D treatment planning system manufactured by Varian Medical Systems. In patients with clinically negative neck the gross tumor volume (GTV) was represented by the gross tumor volume of the primary tumor (GTV70) only and defined as any visible tumor revealed on imaging studies and/or physical examination. In patients with clinically positive neck the GTV70 was an union of GTV70 and GTVn70. The GTVn70 was defined as the gross nodal disease revealed on imaging studies and/or physical examination. Neck lymph nodes were considered metastatic when their smallest axis diameter was greater than 1.0 cm. Definitions of clinical target volume (CTV) are shown in Table 1.

<table>
<thead>
<tr>
<th>Target</th>
<th>Target delineation</th>
</tr>
</thead>
<tbody>
<tr>
<td>CTVt50</td>
<td>GTVt70 plus a margin of 1.0-2.0 cm for the potential microscopic extension of the disease</td>
</tr>
<tr>
<td>CTVn50 in patients with clinically negative neck</td>
<td>Nodal regions in the neck at levels II–IV</td>
</tr>
<tr>
<td>CTVn50 in patients with clinically positive neck</td>
<td>GTVn70 with a margin of 0.5-1.0 cm, nodal regions in the neck at levels I-V, retropharyngeal lymph nodes plus retrostyloid space if positive lymph node(s) in level Ib plus supraclavicular fossa if positive lymph node(s) in levels IV or V</td>
</tr>
<tr>
<td>CTV50</td>
<td>Created by integration of CTVt50 and CTVn50</td>
</tr>
</tbody>
</table>

The planning target volumes were PTV50 and PTV70. The PTV50 provided a margin of 0.5 cm around CTV50. If there were no positive lymph nodes in the neck, the PTV70 encompassed the GTV70 plus a 0.5 cm margin. In patients with nodal disease, PTV70 was obtained by adding a margin of 0.5 cm around GTV70. Radiotherapy was delivered on linear accelerator using photons with beam qualities of 6 MV and 15 MV and electrons with energies 9-16 MeV. Conventional fractionation was used with a daily dose of 2.0 Gy, 5 times per week.

2.1.2. Chemotherapy

Chemotherapy consisted of weekly cisplatin (30 mg/m2) given concomitantly with radiation started at the first day of radiotherapy. Hydration and antiemetics were delivered according to standards of care. The full blood count and biochemical analysis of serum urea and creatinine were checked every week.
2.2. Response assessment
Evaluation of tumor response was performed three months after completion of chemoradiotherapy by physical examination, CT or MRI, and fiberoptic endoscopy. An examination under anesthetic and biopsies were performed in the event of clinical, endoscopic or radiological abnormalities. Response to treatment was documented by the World Health Organization (WHO) response grading system [25].

2.3. Treatment toxicity assessment
During treatment, the patients were examined on a weekly basis. After completion of therapy, each patient was followed up clinically after 4-6 weeks to assess acute toxicity. Acute radiotherapy-related toxicities were assessed according to the Acute Radiation Morbidity Scoring Criteria of the RTOG [26]. Chemotherapy-related toxicities were assessed according to the World Health Organization (WHO) criteria [25]. Late radiotherapy-related toxicities were evaluated according to the scales of the EORTC/RTOG [26] and were recorded starting at 4 months after treatment completion.

2.4. Follow-Up
All patients were followed up every month over the first year, every other month in the second year, every 3 to 6 months in the third through the fifth years after treatment, and every 12 months thereafter. A physical examination and fiberoptic endoscopy, or indirect mirror exam were performed during each follow-up examination. Baseline CT and/or MRI of the neck were done every 6 months over the first 2 years. Biopsy was performed in order to obtain histological proof of any lesion clinically suspicious for recurrent disease. Chest radiography and ultrasonography of the liver were performed each year.

2.5. Statistical analysis
Statistical end points of this study were LRRFS, DFS, and OS. LRRFS for patients who achieved complete composite response to CCRT was measured from the first day of treatment to the date of reappearance of disease either at the primary site and/or regional lymph nodes, or until the day of the last follow-up. For patients initially staged as N0 who manifested complete primary response to treatment, LRRFS was calculated from the date of treatment beginning until the date when appearance of metastatic lymph node(s) in the neck and/or reappearance of disease at the primary site were first reported, or to the last follow-up date. For patients with persistent primary and/or nodal disease LRRFS was measured from the first day of treatment to the date of the first follow-up visit. DFS was calculated from the date of commencement of treatment to the date when local, regional, locoregional or distant failure was first recorded or, in the case of local and/or regional persistent disease, to the date of first follow-up visit. OS was measured from the start date of treatment to the date of the last follow-up or to the date of death from any cause. LRRFS, DFS and OS were calculated using the method of Kaplan-Meier [27].

3. RESULTS

3.1. Patient and Tumor Characteristics
Summaries of baseline patient and tumor characteristics are provided in Table 2. The median age for the entire group was 58.5 years (range, 36-69 years), 88.9% were male, and 69.4% had an Eastern Cooperative Oncology Group (ECOG) performance status score of 0. The predominant T and N stage was T3 (69.4%) and N2 (36.1%), and well and moderate histological differentiation of the tumor were equally represented (38.9%). Stage IVA disease was recognized in 58.3% of the patients.

Table 2. Baseline patient and tumor characteristics (n = 36)

<table>
<thead>
<tr>
<th>Characteristics</th>
<th>Number of patients (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Patient characteristics</strong></td>
<td></td>
</tr>
<tr>
<td>Gender:</td>
<td></td>
</tr>
<tr>
<td>male</td>
<td>32 (88.9%)</td>
</tr>
<tr>
<td>female</td>
<td>4 (11.1%)</td>
</tr>
<tr>
<td>ECOG performance status:</td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>25 (69.4%)</td>
</tr>
<tr>
<td>1</td>
<td>11 (30.6%)</td>
</tr>
<tr>
<td><strong>Tumor characteristics</strong></td>
<td></td>
</tr>
<tr>
<td>T stage:</td>
<td></td>
</tr>
<tr>
<td>T3</td>
<td>25 (69.4%)</td>
</tr>
<tr>
<td>T4</td>
<td>11 (30.6%)</td>
</tr>
<tr>
<td>N stage:</td>
<td></td>
</tr>
<tr>
<td>N0</td>
<td>11 (30.6%)</td>
</tr>
<tr>
<td>N1</td>
<td>8 (22.2%)</td>
</tr>
<tr>
<td>N2</td>
<td>13 (36.1%)</td>
</tr>
<tr>
<td>N3</td>
<td>4 (11.1%)</td>
</tr>
<tr>
<td>Stage:</td>
<td></td>
</tr>
<tr>
<td>III</td>
<td>11 (30.6%)</td>
</tr>
<tr>
<td>IVA</td>
<td>21 (58.3%)</td>
</tr>
<tr>
<td>IVB</td>
<td>4 (11.1%)</td>
</tr>
<tr>
<td>Histological differentiation:</td>
<td></td>
</tr>
<tr>
<td>well</td>
<td>14 (38.9%)</td>
</tr>
<tr>
<td>moderate</td>
<td>14 (38.9%)</td>
</tr>
<tr>
<td>poor</td>
<td>8 (22.2%)</td>
</tr>
</tbody>
</table>

ECOG, Eastern Cooperative Oncology Group

3.2. Compliance with treatment
The prescribed total radiotherapy dose of 70 Gy was given in all the patients (100%). Twenty nine patients (80.6%) completed the radiotherapy course in a period of time ≤ 7 weeks. Twenty two patients (61.1%) received all seven cycles of concurrent cisplatin. The mean total dose of cisplatin given was 192 mg/m2 ± 14.8 SD.

3.3. Response to treatment
At 3 months post treatment assessment 26 patients (72.2%) achieved complete response at the primary
site (Table 3). Complete response of the nodal disease was seen in 16 out of 25 patients presented with metastatic lymph node(s) in the neck (64.0%) (Table 3). There was no salvage neck dissection performed for residual neck disease in the remaining 9 patients (36.0%) who achieved a partial response of the nodal disease. Complete composite response was achieved in 25 patients (69.4%) (Table 3).

Table 3. Tumor responses assessed clinically and radiologically 3 months after completion of concurrent chemoradiotherapy

<table>
<thead>
<tr>
<th>Response to treatment</th>
<th>Number of patients (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Complete response of the primary tumour</td>
<td>26/36 (72.2%)</td>
</tr>
<tr>
<td>Partial response of the primary tumour</td>
<td>10/36 (27.8%)</td>
</tr>
<tr>
<td>Complete response of the nodal disease</td>
<td>16/25 (64.0%)</td>
</tr>
<tr>
<td>Partial response of the nodal disease</td>
<td>9/25 (36.0%)</td>
</tr>
<tr>
<td>Complete composite response</td>
<td>25/36 (69.4%)</td>
</tr>
<tr>
<td>Partial composite response</td>
<td>11/36 (30.6%)</td>
</tr>
</tbody>
</table>

3.4. Survival outcomes

The median follow-up of all patients was 20.5 months (range, 5 to 90 months). Nine patients (25%) remain alive, with a median follow-up of 59 months (range, 30 to 90 months). In those 25 patients who achieved complete composite response, treatment failure occurred in 15 patients (60.0%). The site of the primary tumor was the most common location of isolated recurrence noted in six patients. An isolated regional recurrence was present in only one patient, while four patients developed locoregional recurrence. Distant tumor failure was observed in six patients. Two of these six patients experienced locoregional failure. The sites of distant metastases included the lung, bones and liver. Lung was the distant metastases site in all the patients. One patient had non-small cell lung cancer as a second primary malignancy.

Eleven patients died because of the progression of local, regional, or locoregional persistent disease, six patients died due to a local recurrence, one patient died because of isolated regional recurrence, two patients died due to a recurrence at both primary and nodal site, four patients died after developing distant metastases, and two patients died because of development of locoregional recurrence and distant metastases. One patient died due to a second primary tumor.

At 3 years, the LRRFS rate was 38.8% (Figure 1). Three years DFS rate was 27.8% (Figure 2). The 3-year OS rate was 27.3% (Figure 3). The median duration of LRRFS and DFS was 12 months (range, 3-90 months), and the median duration of OS was 20.5 months (range, 7-90 months).

3.5. Treatment induced toxicity

Acute toxicities are shown in Table 4. The most frequently manifested radiation induced acute mucosal reaction was grade 3 mucositis experienced in 58.3% of patients. The most commonly present skin toxicity was grade 2 acute skin reaction. Grade 2 nausea was the most frequently recognized nonhematological chemotherapy-related toxicity while grade 1 leukopenia was the most commonly seen hematological chemotherapy-related toxicity.

Table 4. Acute toxicity recorded during treatment

<table>
<thead>
<tr>
<th>Toxicity</th>
<th>Grade of reaction (% of 36 patients)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0</td>
</tr>
<tr>
<td>Acute normal tissue reactions according to RTOG criteria</td>
<td></td>
</tr>
<tr>
<td>Organ/Tissue</td>
<td></td>
</tr>
<tr>
<td>Skin</td>
<td>0</td>
</tr>
<tr>
<td>Mucous membrane</td>
<td>0</td>
</tr>
<tr>
<td>Acute toxicity according to WHO criteria</td>
<td></td>
</tr>
<tr>
<td>Nonhematological</td>
<td></td>
</tr>
<tr>
<td>Nausea</td>
<td>44.4</td>
</tr>
<tr>
<td>Vomiting</td>
<td>77.8</td>
</tr>
</tbody>
</table>
The average weight loss during treatment as a percentage of weight on treatment start was 9% with a range between 3.0% and 15.0%.

Late toxicities are shown in Table 5. Xerostomia was the most important late complication with grade 2 late salivary gland toxicity revealed in 72.2% of patients. Grade 2 late mucosal reactions was experienced in 58.3% of patients.

Table 5. Late toxicity recorded following treatment

<table>
<thead>
<tr>
<th>Organ/Tissue</th>
<th>Grade of reaction (% of 36 patients)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
<td>61.1 38.9</td>
</tr>
<tr>
<td>Subcutaneous tissue</td>
<td>61.1 38.9</td>
</tr>
<tr>
<td>Mucous membrane</td>
<td>41.7 58.3</td>
</tr>
<tr>
<td>Salivary gland</td>
<td>27.8 72.2</td>
</tr>
</tbody>
</table>

4. DISCUSSION

CCRT using cisplatin as the chemotherapy agent of choice has been widely adopted as the standard of care for locally-regionally advanced HNSCC [12-14]. The benefit of synchronous administration of radiotherapy and chemotherapy compared with radiotherapy alone in the treatment of advanced oropharyngeal cancer has also been well established in the GORTEC 94-01 study [16, 17].

In our study of CRCT using 3DCRT and concurrent weekly cisplatin, the observed 3-year rates of LRRFS, DFS, and OS were 38.8%, 27.8%, and 27.3%, respectively. Unfortunately, the impressive feature when comparing our results with those achieved in studies using definitive radiotherapy alone in the treatment of advanced tonsillar cancer, is the fact that, despite the combination of chemotherapy and radiotherapy used in our study, the obtained results are lower than those reported by authors who used radiotherapy alone. For example, Poulsen et al. [28], comparing outcomes of primary surgery and definitive radiotherapy in patients with stage III and IV squamous cell carcinoma of the tonsil, reported 5-year locoregional control and OS rate of 73% and 41% in the group treated with radiotherapy alone, respectively. In the study of Lee et al. [29] of 243 patients with squamous cell carcinoma of the tonsillar region treated with radical radiotherapy, the reported 5-year absolute and cause-specific survival rates for stage III were 55% and 85%, respectively, for stage IVA were 35% and 60%, respectively, and for stage IVB were 23% and 38%, respectively. Mendenhall et al. [30], in their retrospective study of definitive radiotherapy for carcinoma of the tonsillar area, reported that 5-year cause-specific survival rates for stage III, IVA, and IVB were 84%, 73%, and 46% respectively.

Regarding the analysis of the role of CCRT, specifically in the treatment of locoregionally advanced tonsillar carcinoma, the reported results in the literature are scarce. In the retrospective study of Koo et al. [31], treatment outcome following definitive chemoradiotherapy versus postoperative radiotherapy in patients with stage III-IV tonsillar cancer was analyzed. The reported rates of 5-year locoregional progression-free survival and OS for patients treated with CCRT were 83% and 76%, respectively. One of the studies exploring the efficacy of chemotherapy, as a part of a multimodality approach in locally advanced stage IV tonsillar cancer is the retrospective study of Prestwich et al. [32] reporting the results of induction chemotherapy followed by cisplatin CCRT. The achieved 3-year locoregional control rate in complete responders following induction chemotherapy was 91%. The authors reported 3-year progression-free survival and OS rate of 75% and 66%, respectively. In the retrospective analysis of therapy in 61 patients with pathologically confirmed tonsillar carcinoma without distant metastasis reported by Wang et al. [33], CCRT was used in 21 out of 45 patients with stage III-IV disease. The 5 years survival for this patient’s category was 36.4%. In the study of Shirazi et al. [34], aimed to review their experience in the management of advanced tonsillar squamous cell carcinoma and to compare treatment outcomes between patients treated with and without surgery to the primary site, 74 patients were retrospectively analyzed. Thirty six of these patients were treated with an organ preservation approach using radiotherapy alone or chemoradiotherapy, and 38 patients were treated with definitive surgery. The rates of 4-year local control and 4-year OS for organ preservation group were 86% and 48%, respectively. All these results of the studies using combined treatment approach with radiotherapy and chemotherapy for advanced tonsillar cancer regarding both LRC and survival are remarkably superior to the results obtained in our study.

In an effort to explain such unexpectedly poor results of 3DCRT given with concomitant weekly cisplatin in patients with advanced tonsillar carcinoma in our study, and taking into account that the presence of large primary tumors (T3-T4) or advanced stage (N2-N3) metastatic neck lymph nodes had a detrimental effect on prognosis, we underline that these results could only be partially attributable to the high percentage of stage IVA (58.3%) and stage IVB (11.1%) disease at admission. Another possible explanation for such disappointing results would be insufficient target coverage of the gross tumor, leading to insufficient response of the primary tumor and metastatic nodes in the neck, and also insufficient target coverage of the subclinical disease, leading to local and/or regional failure in patients with complete composite response.

Regarding the acute toxicity, 58.3% of patients in our study experienced grade 3 mucositis. However, the fact that should be emphasized is that besides the
poor outcome of CCRT in our study, when evaluating late radiotherapy-related toxicities starting at 4 months after treatment completion, we revealed that xerostomia was the most prominent late effect with grade 2 present in 72.2% of patients. Considering the unfruitful rates of locoregional control and survival obtained in our study, and the high incidence of high grade xerostomia as the primary quality-of-life complaint among long-term survivors following radiotherapy in head and neck region, we consider necessary to accentuate the role of IMRT representing an advance in technology that allows the radiation oncologist to “shape” radiation dose profiles around normal structures while fully dosing the tumor and risk nodal regions [35]. IMRT allows exquisite dose conformality with sparing of adjacent organs, and therefore orofaryngeal cancers, including those arising from tonsillar region, represent ideal sites for its application since the tumors often present in close proximity to sensitive normal tissues such as the parotid glands. Early experience with IMRT for oropharyngeal cancers shows encouraging results, providing local or locoregional control ranging between 89% and 98% [22, 36, 37]. In the study of Chao et al. [36], comparing tumor control, acute and late toxicity between IMRT and conventional radiotherapy, 460 patients with oropharyngeal cancer were analyzed. Tonsil primary tumors were present in 260 patients. Most of the patients (74%) had stage III-IV disease. The dosimetric advantage of IMRT resulted in a significant reduction of late salivary toxicity without negative impact on tumor control and disease-free survival. In the study of de Arruda et al. [22] the use of IMRT in 40 patients with histologically confirmed cancer of the oropharynx resulted with 2-year locoregional control and OS rate of 98%. The authors of this study conducted in the Memorial Sloan-Kettering Cancer Center reported 33% grade 2 xerostomia at follow-up of nine months or greater. The University of California-San Francisco experience with IMRT and concurrent chemotherapy for stage III and IV oropharyngeal carcinoma showed excellent locoregional control rates and grade 2 xerostomia at follow-up of two years or more reported in 22% of patients [23]. In the retrospective study of Clavel et al. [38] comparing the toxicity and efficacy of IMRT vs. conventional radiotherapy in 249 patients with locally advanced oropharyngeal cancer, the authors reported that IMRT was associated with favorable locoregional control and survival rates with less xerostomia and acute dermatitis than conventional radiotherapy, when both were given concurrently with chemotherapy.

5. CONCLUSION

Taking into account the low 3-year survival rates observed in our study and the high percentage of grade 2 xerostomia, it can be concluded that in the future, instead of 3DCRT with concurrent chemotherapy, intensity modulated radiation therapy (IMRT) in combination with concomitant chemotherapy should be considered a fully recommendable treatment option for advanced tonsillar carcinoma in order to improve locoregional control and survival with simultaneous reduction of late salivary toxicity and therefore prevention of late xerostomia.

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TREATMENT PLANNING SYSTEM COMMISSIONING OF THE ECLIPSE PBC DOSE CALCULATION ALGORITHM FOR THE VARIAN CLINAC iX S/N 5052

Dushko Lukarski¹, Dragan Nikolovski¹

¹ University Clinic of Radiotherapy and Oncology, Vodnjanska 17, 1000 Skopje, Macedonia

Abstract – The commissioning of the Treatment Planning System (TPS) is an important part of the commissioning of a new linear accelerator (linac). In this work, we evaluated the performance of the Pencil Beam Convolution (PBC) algorithm configured for the new Varian Clinac iX (S/N 5052) at the University Clinic of Radiotherapy and Oncology in Skopje. The evaluation was performed in two stages. In the first stage, we used a workspace of the TPS itself, called “Beam Analysis”, in which the system itself calculates the depth dose and profile curves for a water phantom and compares them with those measured during the commissioning of the accelerator. In the second stage, we created, calculated and irradiated 9 test plans on a polystyrene phantom “OPERA” and measured the dose in a point with a system for absolute dosimetry and then compared the measurements with the calculations. In both stages, the results of the comparison were below 3%, in most clinically relevant cases below 2%, which indicates that the PBC algorithm can safely be commissioned for clinical use.

Keywords – TPS commissioning, PBC algorithm, Eclipse TPS

1. INTRODUCTION

The modern radiation therapy treatment planning is a complex process. One of the key components of the process, which only recently started to receive the due attention, is the computerized system for the planning of the treatment. After some serious accidents worldwide, the International Atomic Energy Commission (IAEA) developed series of documents [1, 2] containing general and more detailed recommendations concerning the commissioning of the treatment planning systems (TPS).

At the University Clinic of Radiotherapy and Oncology in Skopje, Macedonia, a new linear accelerator Varian Clinac iX S/N 5052 was recently installed and the commissioning beam data measurements were performed. The measured beam data was imported into the existing treatment planning system at the Clinic – VMS Eclipse ver.10 and with this data, the Pencil Beam Convolution (PBC) treatment planning algorithm was configured. This work reports the performed measurements and the obtained results in the process of the commissioning of this algorithm for clinical use.

2. MATERIALS AND METHODS

The commissioning of the PBC algorithm of the Eclipse TPS was performed in two stages.

In the first stage, we used a specific workspace of the TPS itself, called Beam Analysis [3], in which the system calculates the depth doses and the beam profiles on a virtual water phantom by using the configured algorithm, and compares them with the measured ones.

In the second stage, we used a polystyrene phantom called “OPERA” to evaluate the behavior of the calculation model by performing measurements with ionization chamber and comparing the calculated and measured results.

2.1. Evaluation based on “Beam Analysis” Workspace

As stated before, the workspace “Beam Analysis” is an integral part of the TPS Eclipse. In it, the TPS creates a virtual water phantom and calculates the depth dose curves and the beam profiles and then compares them with those measured during the commissioning of the linac. The depth dose curves were normalized in the maximum of the curve, while...
the profiles were normalized in the center of each profile.

For the depth dose curves, for each field size, the system provides the following comparison results: depth difference in millimeters at maximum dose, depth difference in millimeters at 50% dose, dose difference in percent at 100 mm depth and dose difference in percent at 200 mm depth. In addition, we evaluated the curve that gives the dose difference at different depth and from this curve we established the maximal dose difference between the calculated and the measured depth dose curve.

For the profiles, the system provides the distances between certain dose levels of the measured and calculated profile (ex. distance between the points of the measured and calculated profile where the dose is 80% of the dose at the central axis). We believe that it is even more important to evaluate the maximal dose difference in the flattened area of the profiles. Therefore, for every field size and depth we created an excel worksheet in which we imported the measured and the calculated results and evaluated the maximal dose difference in the flattened area of the profiles. The flattened area of the profile is defined as the central 80% of the field width. The field width is the distance between points of 50% dose (Fig. 1).

![Fig. 1 – Definition of flattened area of a beam profile](image)

2.2. Evaluation based on ionization chamber measurements

For the second stage of the evaluation we used the polystyrene phantom “OPERA” (Fig.2).

We made CT scans of the phantom in three positions (Fig.3), imported the scans in the TPS and created 9 treatment plans. All treatment plans were calculated by using the “No normalization” option in the “Plan Normalization Options” [4, 5]. We irradiated the plans and measured the dose with a Farmer type chamber in a specified point for each plan. The dosimetric system that we used was Scanditronix Welhoffer system consisting of a Farmer type chamber FC-65G and electrometer Dose1. The measurements were corrected for the daily deviation in the output of the linac and for the attenuation of the treatment couch top.

2.2.1. Test 1 – Testing calculation for reference conditions

For this test, the CT shown on Fig.3a was used. The phantom was positioned in a setup where the source to phantom distance (SPD) was 100 cm. Two fields were used - one with 6 MV photons and another with 15 MV photons. For both fields the field size was 10x10 cm², the gantry and collimator angles were 0°. The field weight was such that by each of the fields 100 MU were delivered. The measurement point is marked with the green cross on Fig.3a. It is a point at 5 cm depth and for this point the calculation and the measurement were compared.

![Fig. 2 – “OPERA” phantom](image)

![Fig. 3 – CT scans used in different plans](image)

2.2.2. Test 2 – Testing calculation in case of a lack of scattering for a tangential field

For this test, the CT shown on Fig.3b was used. The phantom was positioned in an isocentric setup with the isocenter as shown on Fig.3b – marked with the green cross. Two rectangular fields were used – one with 6 MV photons and another with 15 MV photons. For both fields the field size was 10x15 cm², the gantry angle was 270° and the collimator angle was 90°. Both fields were with dynamic wedges – Enhanced Dynamic Wedge 60° (EDW60). The prescribed daily dose was 2 Gy and the weight of both fields was equal. The dose was measured at the isocenter (green cross on Fig.3b).
2.2.3. Test 3 – Testing automatic margin function and customized blocking with MLC

A target contour was drawn on the CT shown on Fig.3c and automatically expanded. The phantom was positioned in a setup where the SPD = 100 cm. Two identical fields with MLC (one with 6MV one with 15MV) were fitted automatically to the expanded target structure with a circular margin of 0.7 mm. The gantry and collimator angles were 0°. The prescribed daily dose was 2 Gy and the field weights were such that 40% of the dose was delivered with the 15 MV field while 60% of the dose was delivered with the 6 MV field. That yielded 103 MU for the 15 MV field and 101 MU for the 6 MV field. The point where we compared the measurement and the calculation was at 10 cm depth (marked with the green cross on Fig. 3c). In this test, the two inhomogeneties of the phantom (the air filled space and the cork filled space) were in the path of the beams, so their influence on the MU calculation could be evaluated.

2.2.4. Test 4 – Testing calculation in case of a significant blocking of the field corners

For this test, the CT shown on Fig.3a was used. The phantom was positioned in an isocentric setup, with the isocenter as shown on Fig.3a – marked with the green cross. This was also the measurement point. Two fields were used, one with 6 MV and another with 15 MV. For both fields the gantry angle was 0°, and the collimator angle was 45°. The field size defined by the collimator jaws was 14x14 cm² and the four field corners were blocked with the MLC leaving an opening of 10x10 cm² (Fig.4).

The prescribed daily dose was 2 Gy and the weight of both fields was equal.

2.2.5. Test 5 – Testing calculation in case of an oblique incidence with irregular field and blocking of the center of the field

For this test, the CT shown on Fig.3b was used. The phantom was positioned in an isocentric setup, with the isocenter shown on Fig.3b – marked with the red cross. The measurement point is marked with the green cross on the same figure. It is 5 cm above the isocenter. Two fields were used, one with 6 MV and another with 15 MV. For both fields the gantry angle was 320°, and the collimator angle was 90°. The field size defined by the collimator jaws was 12x13 cm² and an L-shaped field was created by blocking off 5x8 cm² of the field (Fig.5).

On Fig.5 the isocenter position is shown with the red cross and the measurement position is shown with the green cross. The prescribed daily dose was 2 Gy and the weight of both fields was equal.

2.2.6. Test 6 – Testing calculation in case of a four field box

For this test, the CT shown on Fig.3a was used. The phantom was positioned in an isocentric setup, with the isocenter as shown on Fig.3a – marked with the green cross. This was also the measurement point. A four field box test plan was created with 15 MV fields and gantry angles 0°, 90°, 180° and 270°. For each field the collimator angle was 0° and a MLC was placed with an orientation typical for most clinical cases that employ this technique. The prescribed daily dose was 2 Gy and the weights of all four fields were equal.

2.2.7. Test 7 – Testing calculation in case of a typical plan for irradiation of rectal cancer

This was a test plan for evaluating the irradiation of a typical rectal cancer case. For this test, the CT shown on Fig.3a was used. The phantom was positioned in an isocentric setup, with the isocenter as shown on Fig.3a – marked with the green cross. This was also the measurement point. A three field test plan was created with prescribed daily dose of 2 Gy. The gantry angles of the fields were 0°, 90° and 270°. For each of the fields the collimator angle was 90° and a MLC was placed with an orientation typical for most clinical cases that employ this technique. The field with gantry angle 0° was with 6 MV, and the other two fields were with 15 MV and EDW60. The weight of the field with gantry 0° was 60%, while each of the other two fields delivered 20% of the dose.

2.2.8. Test 8 – Testing calculation in case of a typical plan for irradiation of brain tumors

This was a test plan for evaluating the irradiation of a typical brain tumor case. For this test, the CT shown on Fig.3a was used. The phantom was positioned in an isocentric setup, with the isocenter as shown on Fig.3a – marked with the green cross. This was also the measurement point. A three field test plan was created with prescribed daily dose of 2 Gy. For two
of the fields (with 15 MV, EDW10, gantry angles 90° and 270° and collimator angles 0° and 90° respectively) the couch rotation was 0°, while for the third field (6 MV, gantry angle 45°, collimator angle 90°) it was 90°. The weight of the field with a couch angle 90° was 20%, while each of the other two fields delivered 40% of the dose.

2.2.9. Test 9 – Testing calculation in case of a typical plan for irradiation of head and neck cancer

This was a test plan for evaluating the irradiation of a typical head and neck cancer case. For this test, the CT shown on Fig.3a was used. The phantom was positioned in an isocentric setup, with the isocenter as shown on Fig.3a – marked with the green cross. This was also the measurement point. A typical four field test plan was created with prescribed daily dose of 2 Gy. The four fields are described in Table 1.

Table 1. Field description for head and neck case

<table>
<thead>
<tr>
<th>Gantry angle (°)</th>
<th>Coll. angle (°)</th>
<th>Nom. En. (MV)</th>
<th>Weight (%)</th>
<th>EDW</th>
</tr>
</thead>
<tbody>
<tr>
<td>140</td>
<td>0</td>
<td>6</td>
<td>12.5</td>
<td>30</td>
</tr>
<tr>
<td>60</td>
<td>90</td>
<td>6</td>
<td>37.5</td>
<td>30</td>
</tr>
<tr>
<td>300</td>
<td>90</td>
<td>6</td>
<td>37.5</td>
<td>30</td>
</tr>
<tr>
<td>220</td>
<td>0</td>
<td>6</td>
<td>12.5</td>
<td>30</td>
</tr>
</tbody>
</table>

For each of the fields a MLC was placed with an orientation typical for most clinical cases that employ this technique.

3. RESULTS

3.1. Evaluation based on “Beam Analysis”

Workshop

In Tables 2 and 3, we present the results from the comparison of the measured and the calculated depth dose curves in the “Beam Analysis” workspace of “Eclipse” for 6 MV and 15 MV photons, respectively.

Table 2. Difference between calculated and measured depth dose curve for 6 MV photons

<table>
<thead>
<tr>
<th>Square field size (cm)</th>
<th>Max. dose differ. (%)</th>
<th>Depth differ. (mm) at max. dose</th>
<th>Depth differ. (mm) at 50% dose</th>
<th>Dose differ. (%) at 100 mm depth</th>
<th>Dose differ. (%) at 200 mm depth</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>0.50</td>
<td>0.05</td>
<td>0.15</td>
<td>0.30</td>
<td>0.03</td>
</tr>
<tr>
<td>4</td>
<td>0.60</td>
<td>0.05</td>
<td>0.03</td>
<td>0.37</td>
<td>0.01</td>
</tr>
<tr>
<td>6</td>
<td>0.50</td>
<td>0.05</td>
<td>0.10</td>
<td>0.30</td>
<td>0.01</td>
</tr>
<tr>
<td>8</td>
<td>0.52</td>
<td>0.05</td>
<td>0.15</td>
<td>0.32</td>
<td>0.07</td>
</tr>
<tr>
<td>10</td>
<td>0.47</td>
<td>0.05</td>
<td>0.12</td>
<td>0.31</td>
<td>0.01</td>
</tr>
<tr>
<td>12</td>
<td>0.50</td>
<td>0.05</td>
<td>0.10</td>
<td>0.32</td>
<td>0.01</td>
</tr>
<tr>
<td>15</td>
<td>0.40</td>
<td>0.05</td>
<td>0.15</td>
<td>0.29</td>
<td>0.04</td>
</tr>
<tr>
<td>20</td>
<td>0.41</td>
<td>0.05</td>
<td>0.02</td>
<td>0.31</td>
<td>0.04</td>
</tr>
<tr>
<td>25</td>
<td>0.42</td>
<td>0.05</td>
<td>0.15</td>
<td>0.29</td>
<td>0.03</td>
</tr>
<tr>
<td>30</td>
<td>0.40</td>
<td>0.05</td>
<td>0.26</td>
<td>0.20</td>
<td>0.05</td>
</tr>
<tr>
<td>35</td>
<td>0.37</td>
<td>0.05</td>
<td>0.13</td>
<td>0.22</td>
<td>0.03</td>
</tr>
<tr>
<td>40</td>
<td>0.33</td>
<td>0.05</td>
<td>0.04</td>
<td>0.22</td>
<td>0.03</td>
</tr>
</tbody>
</table>

From the Tables 2 and 3, it can be seen that the differences in the depth dose curves are very small and therefore quite acceptable. For example, the maximal dose difference for 6 MV photon beam is less than 0.6%, while for the 15 MV photon beam it is less than 0.5%.

Concerning the evaluation of the difference of the measured and calculated profiles, the “Beam Analysis” workspace calculates the difference (in mm) in the off-axis distance at different dose levels for the left side and the right side of the profiles. In Table 4, we present the maximal values of these differences for 80% and 50% dose, for all the depths of measurements for the different fields.

Table 4. Maximal off-axis distances differences (mm) at different dose levels for all the depths of measurements for different field sizes

<table>
<thead>
<tr>
<th>Maximal off-axis distance difference of the:</th>
<th>For fields ≤ 30 cm</th>
<th>&gt; 30 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>80% dose at the left side (mm)</td>
<td>0.64</td>
<td>2.0</td>
</tr>
<tr>
<td>80% dose at the right side (mm)</td>
<td>0.72</td>
<td>3.3</td>
</tr>
<tr>
<td>50% at the left side dose (mm)</td>
<td>0.64</td>
<td>1.1</td>
</tr>
<tr>
<td>50% dose at the right side (mm)</td>
<td>0.88</td>
<td>1.3</td>
</tr>
</tbody>
</table>

The values for the fields smaller than 30 cm were found to be quite acceptable. For the fields greater than 30 cm (i.e. 35 cm and 40 cm), because of the size of the fields, we had to use a calculation grid of 0.5 cm (instead of grid size 0.25 cm which we used for smaller fields) which also influenced the obtained results.

Concerning the beam profile differences, in addition to the results provided by the “Beam Analysis” workspace, we performed an evaluation of the dose differences within the flattened area. In Tables 5 and 6, the maximal dose differences (in %), between the calculated and the measured beam profiles are given, for all measured field sizes and depths, within the flattened area of the profiles, for 6 MV and 15 MV photon beams respectively.
Table 5. Maximal dose differences between the calculated and the measured profile within the flattened area (%) for 6 MV photon beam

<table>
<thead>
<tr>
<th>Field size (cm)</th>
<th>Depth (cm)</th>
<th>1.6</th>
<th>5</th>
<th>10</th>
<th>20</th>
<th>30</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>0.45</td>
<td>0.90</td>
<td>0.82</td>
<td>0.33</td>
<td>2.07</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>0.50</td>
<td>1.40</td>
<td>1.41</td>
<td>0.84</td>
<td>1.61</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>0.30</td>
<td>1.17</td>
<td>1.02</td>
<td>0.41</td>
<td>0.4</td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>0.25</td>
<td>0.64</td>
<td>0.9</td>
<td>0.73</td>
<td>0.57</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>0.70</td>
<td>0.51</td>
<td>0.69</td>
<td>1.05</td>
<td>1.95</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>0.40</td>
<td>0.60</td>
<td>0.79</td>
<td>1.42</td>
<td>2.06</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>0.61</td>
<td>0.68</td>
<td>0.64</td>
<td>1.65</td>
<td>2.59</td>
<td></td>
</tr>
<tr>
<td>20</td>
<td>0.41</td>
<td>0.87</td>
<td>0.86</td>
<td>1.74</td>
<td>2.79</td>
<td></td>
</tr>
<tr>
<td>25</td>
<td>0.49</td>
<td>0.67</td>
<td>0.62</td>
<td>0.72</td>
<td>1.96</td>
<td></td>
</tr>
<tr>
<td>30</td>
<td>0.49</td>
<td>0.67</td>
<td>0.62</td>
<td>0.72</td>
<td>1.96</td>
<td></td>
</tr>
<tr>
<td>35</td>
<td>0.29</td>
<td>0.62</td>
<td>0.82</td>
<td>0.42</td>
<td>1.05</td>
<td></td>
</tr>
<tr>
<td>40</td>
<td>0.78</td>
<td>1.22</td>
<td>1.14</td>
<td>0.96</td>
<td>0.66</td>
<td></td>
</tr>
</tbody>
</table>

Table 6. Maximal dose differences between the calculated and the measured profile within the flattened area (%) for 15 MV photon beam

<table>
<thead>
<tr>
<th>Field size (cm)</th>
<th>Depth (cm)</th>
<th>1.6</th>
<th>5</th>
<th>10</th>
<th>20</th>
<th>30</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>0.30</td>
<td>0.27</td>
<td>0.53</td>
<td>1.40</td>
<td>2.41</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>0.89</td>
<td>0.94</td>
<td>0.52</td>
<td>2.05</td>
<td>1.27</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>0.75</td>
<td>0.73</td>
<td>0.98</td>
<td>0.94</td>
<td>0.87</td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>0.60</td>
<td>0.89</td>
<td>1.09</td>
<td>0.88</td>
<td>0.76</td>
<td></td>
</tr>
<tr>
<td>10</td>
<td>0.80</td>
<td>0.84</td>
<td>0.94</td>
<td>0.80</td>
<td>1.00</td>
<td></td>
</tr>
<tr>
<td>12</td>
<td>0.90</td>
<td>1.10</td>
<td>1.00</td>
<td>0.69</td>
<td>1.40</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>0.92</td>
<td>1.10</td>
<td>1.21</td>
<td>1.38</td>
<td>1.80</td>
<td></td>
</tr>
<tr>
<td>20</td>
<td>1.29</td>
<td>1.25</td>
<td>0.95</td>
<td>1.39</td>
<td>2.46</td>
<td></td>
</tr>
<tr>
<td>25</td>
<td>1.49</td>
<td>1.13</td>
<td>1.10</td>
<td>1.49</td>
<td>2.36</td>
<td></td>
</tr>
<tr>
<td>30</td>
<td>1.37</td>
<td>1.02</td>
<td>1.11</td>
<td>1.44</td>
<td>1.99</td>
<td></td>
</tr>
<tr>
<td>35</td>
<td>0.86</td>
<td>1.02</td>
<td>0.95</td>
<td>0.82</td>
<td>1.14</td>
<td></td>
</tr>
<tr>
<td>40</td>
<td>1.26</td>
<td>1.27</td>
<td>1.21</td>
<td>1.04</td>
<td>0.82</td>
<td></td>
</tr>
</tbody>
</table>

Table 7. Results of the tests performed with the “OPERA” phantom

<table>
<thead>
<tr>
<th>Test No.</th>
<th>Field</th>
<th>MU</th>
<th>Calculated Dose (cGy)</th>
<th>Measured dose (cGy)</th>
<th>Relative Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>6 MV</td>
<td>100</td>
<td>87.1</td>
<td>86.90</td>
<td>-0.23</td>
</tr>
<tr>
<td>2</td>
<td>15 MV</td>
<td>100</td>
<td>95.2</td>
<td>95.46</td>
<td>0.27</td>
</tr>
<tr>
<td>3</td>
<td>6 MV</td>
<td>183</td>
<td>100.1</td>
<td>98.54</td>
<td>-1.58</td>
</tr>
<tr>
<td>4</td>
<td>15 MV</td>
<td>147</td>
<td>100.6</td>
<td>99.08</td>
<td>-1.53</td>
</tr>
<tr>
<td>5</td>
<td>6 MV</td>
<td>101</td>
<td>66.7</td>
<td>66.26</td>
<td>-0.66</td>
</tr>
<tr>
<td>6</td>
<td>15 MV</td>
<td>103</td>
<td>78.6</td>
<td>78.70</td>
<td>0.13</td>
</tr>
<tr>
<td>7</td>
<td>6 MV</td>
<td>101</td>
<td>98.3</td>
<td>97.99</td>
<td>-0.32</td>
</tr>
<tr>
<td>8</td>
<td>15 MV</td>
<td>94</td>
<td>99.1</td>
<td>99.35</td>
<td>0.25</td>
</tr>
<tr>
<td>9</td>
<td>6 MV</td>
<td>127</td>
<td>126.6</td>
<td>125.18</td>
<td>-1.13</td>
</tr>
<tr>
<td>10</td>
<td>15 MV</td>
<td>109</td>
<td>122.2</td>
<td>120.61</td>
<td>-1.32</td>
</tr>
<tr>
<td>11</td>
<td>g=180</td>
<td>52</td>
<td>49.9</td>
<td>49.71</td>
<td>-0.38</td>
</tr>
<tr>
<td>12</td>
<td>g=90</td>
<td>52</td>
<td>49.7</td>
<td>49.39</td>
<td>-0.63</td>
</tr>
<tr>
<td>13</td>
<td>g=0</td>
<td>47</td>
<td>49.9</td>
<td>50.26</td>
<td>0.72</td>
</tr>
<tr>
<td>14</td>
<td>g=270</td>
<td>52</td>
<td>49.7</td>
<td>49.5</td>
<td>-0.40</td>
</tr>
<tr>
<td>Sum</td>
<td></td>
<td>199.2</td>
<td>198.86</td>
<td></td>
<td>-0.17</td>
</tr>
<tr>
<td>15</td>
<td>g=90</td>
<td>70</td>
<td>40.4</td>
<td>40.22</td>
<td>-0.45</td>
</tr>
<tr>
<td>16</td>
<td>g=0</td>
<td>121</td>
<td>119.6</td>
<td>119.72</td>
<td>-0.10</td>
</tr>
<tr>
<td>17</td>
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<td>69</td>
<td>40.4</td>
<td>39.31</td>
<td>-2.77</td>
</tr>
<tr>
<td>Sum</td>
<td></td>
<td>200.4</td>
<td>199.25</td>
<td></td>
<td>-0.58</td>
</tr>
<tr>
<td>18</td>
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<td>95</td>
<td>80.1</td>
<td>78.93</td>
<td>-1.48</td>
</tr>
<tr>
<td>19</td>
<td>g=90</td>
<td>89</td>
<td>80.7</td>
<td>79.02</td>
<td>-1.37</td>
</tr>
<tr>
<td>20</td>
<td>couch=90</td>
<td>43</td>
<td>39.8</td>
<td>39.62</td>
<td>-0.45</td>
</tr>
<tr>
<td>Sum</td>
<td></td>
<td>200.0</td>
<td>197.57</td>
<td></td>
<td>-1.23</td>
</tr>
<tr>
<td>21</td>
<td>g=140</td>
<td>44</td>
<td>24.6</td>
<td>24.26</td>
<td>-1.4</td>
</tr>
<tr>
<td>22</td>
<td>g=60</td>
<td>106</td>
<td>74.3</td>
<td>72.98</td>
<td>-1.81</td>
</tr>
<tr>
<td>23</td>
<td>g=300</td>
<td>107</td>
<td>74.4</td>
<td>73.35</td>
<td>-1.43</td>
</tr>
<tr>
<td>24</td>
<td>g=220</td>
<td>44</td>
<td>24.7</td>
<td>24.17</td>
<td>-2.2</td>
</tr>
<tr>
<td>Sum</td>
<td></td>
<td>198.0</td>
<td>194.76</td>
<td></td>
<td>-1.66</td>
</tr>
</tbody>
</table>

From Tables 4 and 5, it can be seen that for both photon energies, for all depths and field sizes, the maximal dose difference is smaller than 3%, while for the more significant depths (less than 20 cm) the differences are even less than 1.5%. This is an acceptable result and the “Beam Analysis” workspace justifies the clinical use of the PBC algorithm for a three-dimensional conformal radiotherapy treatment planning.

3.2. Evaluation based on ionization chamber measurements

The results from the tests performed with the “OPERA” phantom are given in Table 7. All the measurements were corrected for the daily deviation in the output of the ion. The measurements where the beam traversed through the treatment couch top were also corrected for its attenuation.

The results given in Table 7 are all within 3%, which justifies the clinical use of the PBC algorithm for a three-dimensional conformal radiotherapy treatment planning.

4. CONCLUSION

From the results obtained from the “Beam Analysis” workspace of the Eclipse TPS, the differences between the measured and the calculated depth dose curves were found to be less than 0.6% for 6 MV and 0.5% for 15 MV. For all the depths the dose difference between the beam profiles was found to be less than 3% and for the clinically significant depths even less than 1.5% within the flattened area. The measurements with the “OPERA” phantom confirmed that the difference between the planned dose and the delivered dose will be less than 3%, and for the clinically significant cases usually less than 2%. All these results justify the clinical use of the PBC algorithm for a three-dimensional conformal radiotherapy treatment planning, as it is configured for the Varian Clinac S/N 5052 at the University Clinic of Radiotherapy and Oncology in Skopje, Macedonia.
5. REFERENCES


DOSIMETRIC COMPARISON OF THE LINEAR ACCELERATORS AT THE UNIVERSITY CLINIC OF RADIOTHERAPY AND ONCOLOGY IN SKOPJE, MACEDONIA

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¹ University Clinic of Radiotherapy and Oncology, Vodnjanska 17, 1000 Skopje, Macedonia

Abstract – In radiotherapy practice, for various practical reasons it is important to know whether two or more linear accelerators (linacs) are dosimetrically matched and whether the patient’s treatment can be shifted from one linac to another without reducing the treatment quality. This work presents the data from the dosimetric comparison of the two Varian Clinacs 23EX and one Varian Clinac iX at the University Clinic of Radiotherapy and Oncology in Skopje. Both Percentage Depth Dose (PDD) and Beam Profile (BP) curves were compared for the photon energies (6MV, 15MV) in use at the clinic. The comparison was performed using the IBA OmniPro Accept 7.4™ software. The results from the comparison of the PDD curves showed that in the clinically significant region the dose differences were smaller than 1%. The results from the comparison of the inline and crossline BP curves showed that in the flattened area the dose differences were smaller than 2.5%, while in the penumbra region they were usually between 2% and 8%, but sometimes up to 21%. This suggests that for treatments where the influence of the penumbra region is small, the three linacs may be considered to be dosimetrically matched. For treatments where the influence of the penumbra region is greater, the patient can be switched to another machine only after recalculation of the treatment plan.

Keywords – Linac matching

1. INTRODUCTION

At the University Clinic of Radiotherapy and Oncology in Skopje three linear accelerators (linacs) are in clinical use: two Varian 23EX Clinacs, in use since 2004 (S/N 356 and S/N 357), and one Varian iX Clinac, in use from 2013 (S/N 5052). Occasionally, one of the linacs is out of use (in case of regular or interventional service) and the patient’s treatments cannot be performed on the dedicated linac. Therefore it is useful to know how big the dosimetric differences between the linacs are. If the differences are small, we can consider them as dosimetrically matched and we can incidentally switch the patients from one linac to another without reducing the quality of their treatment.

2. MATERIALS AND METHODS

During the commissioning of the linacs for clinical use, the Percentage Depth Dose (PDD) curves and the Beam Profile (BP) curves were measured [1, 2] with ionization chamber in water. In this work we compared these curves for all three linacs, for both photon energies used (6MV and 15MV). The comparison was performed with the IBA OmniPro Accept 7.4™ software [3]. The comparison was performed in pairs - linac 356 vs. linac 357, linac 356 vs. linac 5052 and linac 357 vs. linac 5052.

2.1. Comparison of the PDD curves

The measurements of the PDD curves were performed by using an ionization chamber in water phantom, starting from chamber depth of 300 mm all the way to the water surface. The PDD curves were measured for square fields with size varying between 30x30 mm² and 400x400 mm². The curves were normalized at the depth of the dose maximum. The comparison of the curves was performed by subtracting the two curves (subtracting the relative dose reading of the two curves for each depth), thus obtaining a new curve called “evaluation” curve – dose difference vs. depth (Fig.1).
In most cases, the evaluation curve had two regions. In the first region, starting from the water surface to a certain depth (which is always inside the buildup region of the PDD curves) the dose difference was greater, while after this depth the dose difference was smaller and more stable. In this work the first region is called the “inside buildup” (IB) region, while the second region, which has much greater clinical significance, is called the “significant” region (Fig.1). When reporting the results, the maximum dose difference in the corresponding region for every measured field size was reported, as well as the depth of the “region meeting point” $d_{mp}$(mm) – the depth where the IB region ends and the “significant” region begins.

2.2. Comparison of the BP curves

The BP curve gives the relative dose measured in a plane orthogonal to the central axis (CAX) of the beam and is a function relative dose vs. distance from the CAX. The BP curves were normalized to the dose at the central axis. In each plane two BP curves were measured - inline and crossline curve. The inline curve was measured in direction gantry-table, while the crossline curve was measured in direction orthogonal to the inline curve direction. The BP curves were measured for square fields with different field sizes (12 field sizes varying between 30x30 mm$^2$ and 400x400 mm$^2$ at five different depths in water ($d_{max}$, 50 mm, 100 mm, 200 mm and 300 mm)

The comparison of the BP curves was done by subtracting the two corresponding curves from the two linacs that were compared (subtracting the relative dose reading of the corresponding inline/crossline curves for each CAX distance), thus obtaining a new curve called “evaluation” curve – dose difference vs. CAX distance (Fig.2). This curve was evaluated in two regions - the flattened area and the 20%-region.

The flattened area is defined by using the field width (FW) parameter (Fig.3). The field width is determined as the distance between the two points on the profile curve where the dose is 50% of the dose at the CAX (Fig.3). The flattened area is determined as FW minus 2 cm for fields smaller than 10 cm and 0.8*FW for fields ≥ 10 cm.

3. RESULTS

3.1. Comparison of the PDD curves

The results from the comparison of the PDD curves are given in Tables 1, 2 and 3 for the pairs 356 vs. 357, 356 vs. 5052 and 357 vs. 5052 respectively.

Table 1. Comparison of the PDD curves for different field sizes, 356 vs. 357 for 6 MV and 15 MV photons

<table>
<thead>
<tr>
<th>Square field size (mm)</th>
<th>Maximum dose differences 356 vs. 357, for 6MV</th>
<th>Maximum dose differences 356 vs. 357, for 15MV</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>19.4</td>
<td>11.4</td>
</tr>
<tr>
<td>30</td>
<td>3.0</td>
<td>11.5</td>
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<tr>
<td>40</td>
<td>2.7</td>
<td>11.4</td>
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<tr>
<td>80</td>
<td>2.3</td>
<td>13.3</td>
</tr>
<tr>
<td>100</td>
<td>0.5</td>
<td>5.5</td>
</tr>
<tr>
<td>120</td>
<td>0.9</td>
<td>6.4</td>
</tr>
<tr>
<td>150</td>
<td>0.8</td>
<td>7.8</td>
</tr>
<tr>
<td>200</td>
<td>0.6</td>
<td>2.0</td>
</tr>
<tr>
<td>250</td>
<td>2.1</td>
<td>9.2</td>
</tr>
<tr>
<td>300</td>
<td>1.2</td>
<td>8.2</td>
</tr>
<tr>
<td>350</td>
<td>1.4</td>
<td>1.6</td>
</tr>
<tr>
<td>400</td>
<td>1.1</td>
<td>7.2</td>
</tr>
</tbody>
</table>

Fig. 1 - Comparison of two PDD curves with representation of the reporting regions

Fig. 2 - Comparison of two BP curves with representation of the reporting regions

Fig. 3 – Definition of flattened area of a beam profile

The 20%-region is defined as the part of the profile curve where the dose is ≥ 20% of the CAX dose. By reporting both the differences in the flattened area and in the 20%-region we are actually reporting the dose inside the flattened area and outside of it, i.e. inside the penumbra region.
From the results presented in Tables 2 and 3, it can be seen that, as expected, the maximum dose differences between the PDD curves in the IB region were found to be bigger than the ones in the “significant” region. In the IB region for 6 MV, for all three pairs of linacs, the maximum dose differences were found to be smaller than 3% and for 15 MV smaller than 4.5%. It can be seen that for 15 MV, in the IB region, the maximum dose differences between 357 and 5052 are much better than in the other two cases and are smaller than 2%.

The clinically more important part of the comparison, the comparison in the “significant” region, showed that, for all cases, the maximum dose differences in this region are smaller than 1%, which justifies considering all three linacs dosimetrically matched, at least when considering the PDD curves, i.e. the nominal energies of the linacs.

### Table 3. Comparison of the PDD curves for different field sizes, 357 vs. 5052 for 6 MV and 15 MV photons

<table>
<thead>
<tr>
<th>Square field size (mm)</th>
<th>IB region (%)</th>
<th>Significant region (%)</th>
<th>d_{eff} (mm)</th>
<th>IB region (%)</th>
<th>Significant region (%)</th>
<th>d_{eff} (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>1.8</td>
<td>5</td>
<td>9.2</td>
<td>1.8</td>
<td>5</td>
<td>9.2</td>
</tr>
<tr>
<td>40</td>
<td>1.2</td>
<td>5</td>
<td>9.3</td>
<td>1.2</td>
<td>5</td>
<td>9.3</td>
</tr>
<tr>
<td>60</td>
<td>1.3</td>
<td>5</td>
<td>9.4</td>
<td>1.3</td>
<td>5</td>
<td>9.4</td>
</tr>
<tr>
<td>80</td>
<td>2.1</td>
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<td>5</td>
<td>9.6</td>
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<tr>
<td>120</td>
<td>1.8</td>
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<tr>
<td>150</td>
<td>2.0</td>
<td>5</td>
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<td>2.0</td>
<td>5</td>
<td>10.2</td>
</tr>
<tr>
<td>200</td>
<td>1.5</td>
<td>5</td>
<td>10.8</td>
<td>1.5</td>
<td>5</td>
<td>10.8</td>
</tr>
</tbody>
</table>

### Table 4. Comparison of the BP curves, range intervals for all field sizes and measurement depths

<table>
<thead>
<tr>
<th>Range of maximum dose difference (%) for different field sizes and measurement depths</th>
<th>flattened area</th>
<th>20%-region</th>
</tr>
</thead>
<tbody>
<tr>
<td>356 vs. 357, 6 MV, inline</td>
<td>0.2 - 1.6</td>
<td>0.9 - 6.5</td>
</tr>
<tr>
<td>356 vs. 357, 6 MV, crossline</td>
<td>0.3 - 2.3</td>
<td>0.8 - 20.7</td>
</tr>
<tr>
<td>356 vs. 357, 15 MV, inline</td>
<td>0.4 - 1.4</td>
<td>0.8 - 5.2</td>
</tr>
<tr>
<td>356 vs. 357, 15 MV, crossline</td>
<td>0.3 - 1.8</td>
<td>0.2 - 20.7</td>
</tr>
<tr>
<td>356 vs. 5052, 6 MV, inline</td>
<td>0.2 - 1.3</td>
<td>1.8 - 11.3</td>
</tr>
<tr>
<td>356 vs. 5052, 6 MV, crossline</td>
<td>0.4 - 1.4</td>
<td>0.8 - 15.6</td>
</tr>
<tr>
<td>356 vs. 5052, 15 MV, inline</td>
<td>0.3 - 2.4</td>
<td>1.5 - 10.1</td>
</tr>
<tr>
<td>356 vs. 5052, 15 MV, crossline</td>
<td>0.2 - 2.1</td>
<td>1.0 - 15.2</td>
</tr>
<tr>
<td>357 vs. 5052, 6 MV, inline</td>
<td>0.2 - 1.3</td>
<td>1.0 - 7.2</td>
</tr>
<tr>
<td>357 vs. 5052, 6 MV, crossline</td>
<td>0.2 - 1.8</td>
<td>1.6 - 13.8</td>
</tr>
<tr>
<td>357 vs. 5052, 15 MV, inline</td>
<td>0.2 - 1.8</td>
<td>1.3 - 7.7</td>
</tr>
<tr>
<td>357 vs. 5052, 15 MV, crossline</td>
<td>0.2 - 1.6</td>
<td>2.1 - 9.8</td>
</tr>
</tbody>
</table>

From the results presented in Table 4, it can be seen that the maximum dose differences, when comparing the BP curves for the different linacs, in the flattened area were found to be smaller than 2.5%, while for the 20%-region, the maximum differences were much bigger, usually between 2% and 8%, but sometimes up to 21%.

It must be stressed that in the flattened area the values for the maximum dose difference were almost always smaller than 1.5% for 6 MV photons and 2% for 15 MV photons. Only for small number of BP curve pairs, the values were between 2% and 2.5%. These curve pairs were mostly inline BP curve pairs from the comparison of linac 356 vs. 5052, for 15 MV.
These results suggest that the three linacs can be considered dosimetrically matched, when irradiating patients with techniques where the influence of the penumbra region is of small importance. This would imply that patients treated with a 3D CRT may incidentally be switched from one to another linac, without recalculation of the treatment plan. However, for patients treated with more advanced techniques (like IMRT), the treatment plans need to be recalculated before transferring the patients to another linac.

4. CONCLUSION

The results presented above show that when comparing the PDD curves of the three linacs, in the clinically significant region, the maximum dose differences were found to be smaller than 1%. For the buildup region, the maximum dose differences go up to 3% for 6 MV and 4.5% for 15 MV photons, but the clinical significance of these results is much smaller.

When comparing the BP curves, in the flattened area which is clinically most significant, the maximum dose differences were always found to be smaller than 2.5%, and in most cases even smaller than 1.5% for 6 MV beams and 2% for 15 MV beams. Outside the flattened area, i.e. in the penumbra region, the maximum dose differences were found to be bigger – in most cases are between 2% and 8%, but sometimes go up to 21%.

From these results, it can be concluded that for the treatment of patients with techniques where the influence of the penumbra region is of small importance, the three linacs can be considered dosimetrically matched. Therefore, for patients treated with a 3D CRT, in incidental situations, the treatment can safely be transferred to another linac without recalculation. For more advanced treatments like IMRT treatments, where the influence of the penumbra region is more important, the patient should be treated on the dedicated linac, or when switching to another linac is necessary, the treatment plan should be recalculated.

5. REFERENCES

NECESSITY OF MEDICAL IMAGING REGISTRATION FOR BRAIN TUMOR RADIOTHERAPY

Sonja Petkovska¹, Slavica Kraleva¹

¹ University Clinic for Radiotherapy and Oncology, Vodnjanska 17, Skopje, Macedonia, pet.sonja@gmail.com

Abstract – Introduction: For brain tumors, the GTV contouring on a CT scan is not precise enough since the soft tissue is hard to be determined. Considering this, the magnetic resonance is the superior method. The goal of this paper is to present the difference in dose distribution if GTV is contoured on a CT scan instead of on an MRI scan.

Materials and methods: All 20 analysed patients had pretreatment CT scan of the head, immobilized with thermoplastic mask. Pre-operative MRI is attached to the patient. Both 3D scans are registered. Two plans are made for each patient. Pr is the real one made according to PTV, where GTV is contoured on MRI-slices and all other volumes (CTV, PTV and OARs) on CT-slices. Pc is a plan for comparison, made according to PTVc. Like all other structures, GTVc is contoured on a CT-scan.

Results: Three parameters are analysed, GTV, CTV dose distribution and conformity index. Analysis results show differences in the volumes of structures GTV and GTVc between 4 and 100 cm³ (Fig. 3), or between 2,3 and 21 mm in diameter if we transform volumes in spheres. The minimal dose coverage of CTV with all plans Pr is above 95 %, except in two cases (93,5 % and 94,5 %). But, only with three Pc plans the dose coverage of CTV is above 90 %. Index conformity values from the analysis show high level of conformation for all Pr plans and for one half of the Pc plans. For one third of the Pc plans, that values are not acceptable. The results presented in this paper show that the contouring of GTV without using MRI in modern radiotherapy results in sub-dosing of the volume of interest; and that multiple increase of uncertainty results in lower local tumour control.

Keywords – brain tumor, dose distribution, conformity index

1. INTRODUCTION

Medical imaging (historically: X-ray, CT, ultrasound, MRI, SPECT, and PET) allow the physician to look inside the body of the patient, without resorting to invasive methods. It is a comfortable and safe diagnostic method for the patient. Today, the radiotherapy treatment of a patient is rarely done without the use of imaging technology. In this paper, we are focusing on CT and MRI images used in radiotherapy for registration of 3D head images from the same subject [1].

1.1. Computerized Tomography and Magnetic Resonance Imaging

Computerized Tomography (CT) scan is the method that involves taking X-ray pictures from various angles and then combining them together to reconstruct the 3D structures in the body [2, 3]. It is far easier to separate tissue other than bone from one another, as compared to simple X-ray scans. However, the soft tissue X-ray absorption is relatively small, and the patient must be exposed to radiation. For a detailed imaging of the anatomy, MRI is the preferred modality.

Magnetic Resonance Imaging, or MRI, is a concept based on the interaction between an external magnetic field and a nucleus that possesses spin. It provides the most detailed anatomical information.

Fig. 1 – CT and MR axial head images
The contrast between the soft tissues is better displayed on MRI, which is why MRI is being used to image the brain and the soft tissue, while CT is used for imaging of bone. Axial head images acquired with CT and MR, respectively, are shown on Fig. 1 for comparison.

1.2. Registration

Registration of two medical images can be defined as a process of finding the geometrical transformation that, when applied to the second image, will align the depicted objects. Because of the difference in intensity values of the same tissue in CT and MR images, multimodality image registration based on comparing intensity values is not trivial. Even more, it is practically impossible to have the patient's head positioned in the scanner exactly the same way as the first time, so the information in the two images will not be spatially aligned [3,4].

2. MATERIALS AND METHODS

Pre-treatment positioning of the patients prepared for radiation therapy is performed on a CT scanner. According to the internal protocol, the part of the patient’s body that includes volume of interest is scanned. The physician delineates tumour or its bed (if it has been surgically removed) on each transversal CT cross section. Presented in 3D, these contours give us the GTV (Gross Tumor Volume). CTV (Clinical Target Volume) is contoured around GTV. That is the volume of interest for radiotherapy. According to the current protocol, the dose between 95% and 107% of the prescribed dose should be delivered in each point of the CTV. To be able to provide it, the treatment plan is made around PTV that includes CTV plus margins (internal and set-up). Additionally, contouring is done on organs with risk.

For the brain tumour patients, the head has been immobilised with thermoplastic face cover mask before the CT scanning. The CT scan thickness through the whole scanning region is 0.5 cm. The patients are in supine position and a CT isocenter is marked on the mask. For this localisation, GTV contouring on CT is not precise enough - since the soft tissue is hard to be determined on a CT scan. Considering this, the magnetic resonance is the superior method. In order to be properly defined, the GTV should be contoured on MRI. Because of this, a pre-surgical MRI is added (is connected, associated, annexed, attached) to the patient. After that, we do a registration of both 3D scans enabling us the random head position during MRI which is matched with the CT scan position that is reference, one with acceptable uncertainties. For the purpose of this paper, a 3D registration package by Varian is used. There are two possibilities, automatic and manual registration. The automatic registration usually provides an acceptable matching of the two 3D scans in the region of interest. If not, the radiation oncologist chooses at least 4 pairs of points with the same anatomical position on both scans. As each plan is defined by 3 points, the forth provides definition of all three plans (XoY, XoZ and YoZ).

![Automatic (a) and manual (b) registration](image)

Fig. 2 – Automatic (a) and manual (b) registration

After registration, the radiation oncologist is contouring GTV on MRI, but all other contours (CTV, PTV, OARs) on CT. The treatment plan (Pr-plan) is made in such a way so that the treatment fields are fitted around PTV with a set-up margin of 0.7 cm.

When MRI, for whatever reason, is not attached to the patient, the GTV has to be contoured on CT. This was the common contouring method in our hospital few years back; since we had no option for MRI – CT registration at the time. The aim of this study is to quantify and discuss the errors made in that case.

Analysis is done on 20 brain tumour (high grade gliomas) patients that had postoperative radiotherapy according to the existing internal procedure. For the purpose of this analysis, before the MRI image is connected to the patient, all the structures (GTVc, CTVc around it and PTVc around it, where „c” stands for comparison) are contoured on CT. Even the GTV is difficult to be recognized, therefore, it is contoured according to the medical (operative) information for tumour size and localization. The treatment plan (Pc-comparison plan) is made in such a way that treatment fields are fitted around PTVc with a set up margin of 0.7 cm. In all 20 patients the volume of interest is far from OARs – lens and eyes. If the volume of interest is close to this OARs than we need to do correction of the treatment fields in both plans. The same correction, however, is impossible because of the two different volumes - PTV and PTVc.

The contouring has been performed by the same radiation oncologist and the plans by the same physicist to avoid uncertainty of human factor – systematic errors. The plans are normalised on the same GLOBAL MAX. What is of interest is, to compare the GTVs dimension, the dose distribution
into CTV according to the plans Pr and Pc, as well as the index conformity.

### 3. RESULTS

#### 3.1. Differences in GTV contouring

Analysis results show differences in the volumes of structures GTV and GTVc between 4 and 100 cm³ (Fig. 3), or between 2, 3 and 21 mm in diameter if we transform volumes into spheres (as in Table 1 below).

**Table 1. GTV and sphere for all analyzed patients**

<table>
<thead>
<tr>
<th>GTV (cm³)</th>
<th>ΔV (cm³)</th>
<th>d (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>plan Pc</td>
<td>plan Pr</td>
<td>plan Pc</td>
</tr>
<tr>
<td>17.7</td>
<td>21.4</td>
<td>3.7</td>
</tr>
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<td>116.6</td>
<td>103.2</td>
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</tr>
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<td>52.0</td>
<td>99.9</td>
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<td>60.0</td>
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<td>89.8</td>
<td>103.9</td>
<td>14.1</td>
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<td>60.9</td>
<td>82.0</td>
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<td>31.3</td>
</tr>
<tr>
<td>62.7</td>
<td>102.3</td>
<td>39.6</td>
</tr>
</tbody>
</table>

Fig. 3 – GTV contoured on CT and MRI and their differences

In 90% of the cases, the GTV is larger (only with two patients the GTVc is larger than the GTV). The information for the dose distribution into CTV are taken from DVH (dose-volume histohram) and dose statistics.

#### 3.2. CTV dose coverage

An individual DVH for CTV is shown below (Fig. 4). There are also dose statistics for both plans Pr and Pc, that give us information about the minimal, the maximal and the mean dose into CTV and PTV. Curves „a“ and „b“ belong to the plans Pr and Pc, respectively.

**Fig. 4 – DVH comparison and dose statistic**

DVH and the dose statistics for all analyzed patients show that the minimal dose coverage of CTV in all Pr plans is above 95 %, except in two cases (93.5 % and 94.5) when CTV and PTV contours are close to the skin. But, only in 3 Pc plans the dose coverage of CTV is above 90 %. Herewith, as it is shown below (Table 2) the Dmean differences into Pr and Pc are negligible.

**Table 2. Doses into CTV**

<table>
<thead>
<tr>
<th>doses into CTV</th>
<th>Dmin (%)</th>
<th>Dmean (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>plan Pc</td>
<td>plan Pr</td>
<td>plan Pc</td>
</tr>
<tr>
<td>13,1</td>
<td>93,5</td>
<td>97,3</td>
</tr>
<tr>
<td>93,40</td>
<td>96,20</td>
<td>101,20</td>
</tr>
<tr>
<td>average</td>
<td>59,06</td>
<td>94,98</td>
</tr>
</tbody>
</table>

#### 3.3. Index conformity

Low CTV coverage means lower PTV coverage, which influences the index conformity. The index conformity is defined in order to determine the quality of conformation. As bigger part of PTV is in 95 % dose, as better quality of conformation is achieved. It is presented as a relation between the volume of the reference dose (VRI) and the target volume(TV).

\[ \text{Conformity index}_{RTOG} = \frac{\text{VRI}}{\text{TV}} \]  

According to the RTOG guidelines, if the conformity index is situated between 1 and 2, the treatment is considered to comply with the treatment plan; an index between 2 and 2.5, or 0.9 and 1, is considered to be a minor violation; and when the index value is less than 0.9 or exceeds 2.5, the protocol violation is considered to be major, but may nevertheless be considered to be acceptable [5, 6]. As it is shown below (Fig.4), the index conformity values for all Pr plans are between 1 to 2 except two (0,98 and 0,96), which means high level of conformation. In spite of
that, in one third of the Pc plans the conformity index values are out of limits - 0.9 and 2.5, or one half of them are between 1 and 2.

![INDEX CONFORMITY](image)

**Fig. 4 – Index conformity for both group of plans**

The worst conformity index values belong to the plans with small GTV and GTVc and when V(GTVc) > V(GTV). The more GTVc and GTV are shifted, the less conformity is obtained.

4. CONCLUSION

Three different parameters important for good brain tumor radiotherapy treatment were analysed in this paper. The GTV location and dimension should be determined as precisely as possible. It influences the CTV and PTV dose distribution which should provide high level of conformity. The results presented in this paper show that the contouring of GTV without using MRI in modern radiotherapy results in sub-dosing of the volume of interest and that multiple increase of uncertainty results in lower local tumour control.

5. REFERENCES


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PORTAL VERIFICATION FOR BREAST CANCER RADIOTHERAPY

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¹ University Clinic for Radiotherapy and oncology, Vodnjanska 17, Skopje, Macedonia, pet.sonja@gmail.com

Abstract – At the University Clinic in Skopje, breast cancer irradiation is being planned and performed by using a mono-isocentrical method, which means that a unique isocenter (IC) for all irradiation fields is used. The goal of this paper is to present the patient’s position in all coordinates before the first treatment session, relative to the position determined during the CT simulation. Deviation of up to 5 mm is allowed. The analysis was made by using a portal verification. Sixty female patients at random selection are reviewed. The matching results show that for each patient deviation exists at least on one axis. The largest deviations are in the longitudinal direction (head-feet) up to 4 mm, mean 1.8 mm. In 60 out of 85 analysed fields, the deviation is towards the head. In lateral direction, median deviation is 1.1 mm and in 65% of the analysed portals those deviations are in medial direction – contralateral breast which can increases the dose in the lung and in the contralateral breast. This deviation for supraclavicular field can increase the dose in the spinal cord. Although these doses are well below the limit, this fact should be taken into account in setting the treatment fields. The final conclusion from the research is that despite of the fact we are dealing with small deviations, in conditions when accuracy in positioning is done with portal, the portal verification needs to be done in the coming weeks of the treatment, not only before the first treatment. This provides information for an intrafractional set-up deviation.

Keywords – isocenter, CT-simulation, portal verification

1. INTRODUCTION

This is the most common type of radiation therapy for women with breast cancer. The radiation is focused from a machine outside the body on the area affected by the cancer. The extent of radiation depends on whether mastectomy or breast –conserving surgery was done and whether or not lymph nodes are involved. The treatment consists of two angled (tangential) beams designed to minimize the dose to the underlying normal lung tissues. A similar approach is used in women treated to the chest wall following mastectomy. In women found to have a lymph node involvement, the radiation is also delivered to the regionally lymph nodes (axillary and supraclavicular region). In these cases, additional beams are mathed to the tangential (breast) fields.

At the University Clinic in Skopje, breast cancer irradiation is being planned and performed by using a mono-isocentrical method, which means that a unique isocenter (IC) for all irradiation fields is used. The treatment is planned to be performed through two tangential fields if only the breast should be irradiated or with three fields if supra and infra-clavicular nodes should be included into the region of interest (two tangential for thorax wall irradiation and supra field for supra and infra-clavicular nodes irradiation). In longitudinal direction (head-feet), the IC is located in the middle of the tangential fields in the first case, and between tangential and supra fields in the second (Fig.1).

Fig. 1 – Supra and Tangential field arranging

When designing tangential fields, the outer limit from the skin is 2 cm measured from projecting furthest point (Fig.1). This is to provide a projection of the breast (or chest wall) being in the radiation field during treatment because of the normal breathing.
2. MATERIALS AND METHODS

The analysis was made for 60 female patients at random selection. During the irradiation the patient lies on a breast board. The pre-treatment positioning and the definition of the treatment isocenter (IC) are performed on a CT simulator, where the region of interest is scanned. The IC location is marked with two tattoos, one at the intersection of transversal and sagital laser and the other in the point of transversal with coronary laser intersection (Fig. 2).

![Fig. 2 – Supra field matching](image)

The treatment plan is often made without changing the isocenter, defined during the CT simulation. Daily patient positioning should provide lasers to pass through tattoos and SSD for each irradiation field to match the default in the plan. In that case, no deviation is expected. But in reality, an ideal match is impossible. Before the first treatment fraction, portal verification, which checks deviation of position on all 3 axes, is mandatory (Fig. 3 and 4).

![Fig. 3 – Tangential field matching](image)

If necessary, the verification procedure can be repeated before the next session or later. Deviation of up to 5 mm is allowed. For deviations greater than 5 mm treatment can not start.

3. RESULTS

Matching results are presented in three axes. Vertical direction represents the Z axis. The deviations in that direction are anterior posterior deviations for the patient. Longitudinal direction represents Y axis or for the patient head-feet direction. Lateral – X axis for the patient is left – right. An individual matching result is shown on Fig. 5 below.

![Fig. 5 – An individual matching result](image)

Analysis shows that for each patient deviation exists at least on one axis. 15 patients – 25%, had deviation in all three axis and 40 patients – 67% had deviations in 2 axes. Analysis results show deviations as on Table 1 and Fig. 6 below.

As it is shown in Table 1, the deviations are minimal in vertical direction (anterior – posterior direction), where the mean deviation is 1 mm. The largest deviations are in the longitudinal direction (head-feet direction), mean 1.8 mm. It is intriguing that only seven had no deviations in that direction. Eighteen deviations are at the feet direction, and the remaining 60 analysed fields (out of 85) had deviation towards the head (Fig. 2 and 3). This is probably result of the steepness of the breast board patients are lying on, as well as the fact that the hand is more relaxed than during a CT scanning done usually two weeks before the treatment. Deviations are not larger than 4 mm, which is acceptable.

![Fig. 4 – Supra field matching results](image)

<table>
<thead>
<tr>
<th>deviation (mm)</th>
<th>max</th>
<th>min</th>
<th>mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>vert</td>
<td>3</td>
<td>-3</td>
<td>1</td>
</tr>
<tr>
<td>lgn</td>
<td>4</td>
<td>-4</td>
<td>1.86</td>
</tr>
<tr>
<td>lat</td>
<td>4</td>
<td>-4</td>
<td>1.12</td>
</tr>
</tbody>
</table>

In lateral direction, the median deviation is 1.1 mm and in 65% of the analysed portals those deviations are in medial direction – contralateral breast. However, the biggest deviation is 4 mm, and in worst case scenario for tangential field with 20 cm length, the projection (screening area) of the irradiation field that will be shifted in medial direction is only 8 cm². In this case, increasing the dose in the lung and in the
contralateral breast would still be acceptable. This deviation for supraclavicular field can increase the dose in the spinal cord. Although this dose is well below the limit, this fact should be taken into account in setting the treatment fields. The supraclavicular treatment field should be designed in such a way that the medial limit is at least 4 mm out of the spinal cord, ensuring that the spinal cord is definitely outside the irradiation field.

Additionally, matching results show that the longituudinally deviations in any patient are not identical to supra and tangential field. It can be as a result of the uncertainties of the portal imager itself. Furthermore, the portal verification is time consuming, and the patient can change her position a little bit between the two portals. It should be also taken into account that to face a radiotherapy is a special psychological moment for the patient, and certainly has an impact on relaxation and position accuracy, especially during the first treatment. But to be able to make a proper conclusion for that, it is necessary to conduct further analyses. It can be a topic for advanced research.

4. CONCLUSION

According to the matching results, there is a good agreement in the position of the breast cancer patient before the first treatment compared to that in the CT simulation. The presented deviations in patients are negligible in clinical practice. They can be attributed to random errors due to the patient’s movement and breathing. Furthermore, more stringent limit (4 mm) can be recommended. To decide to apply this recommendation, more portal verification for each patient should be made. The final conclusion from the research is that despite of the fact we are dealing with small deviations, in conditions when accuracy in positioning is done with portal, the portal verification needs to be done in the coming weeks of the treatment, not only before the first treatment. This provides information for an intrafractional set-up deviation.

5. REFERENCES


SKIN TOXICITY DURING HIPOFRACTIONATED BREAST IRRADIATION IN PATIENT WITH EARLY BREAST CANCER

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Abstract – Radiotherapy is an important component in the treatment of breast cancer. (1) Many women with an early stage of breast cancer are candidates for a breast conservation therapy, which combines both conservative surgery and radiotherapy. (2) According to the data from some series, an estimated 90% of the patients treated with radiotherapy for breast cancer will develop a degree of radiation-induced dermatitis. (3) The severity of the skin reactions during and following the breast irradiation is influenced by both treatment-related and patient-related factors. The treatment-related factors include the fraction size (the dose delivered with each treatment), the total dose delivered, the volume of tissue treated, the type of radiation (4) and the addition of chemotherapy. (5) The patient-related factors include breast size, smoking, axillary lymphoucle drainage before treatment, age, and infection of the surgical wound. (6) A hypofractionation radiotherapy is alternative for a standard fractionation radiotherapy for women with early stage of breast cancer after conservative surgery. The aim of the study was to analyse the acute skin reactions during a hypofractionated radiotherapy in patients with early breast cancer at our institution.

Materials and methods: Twenty patients with early stage of breast cancer (Stadium I and II) and conservative surgery (quadrantectomy of breast with ipsilateral axillary dissection) were analysed. The patients were treated with 6MV x rays on LINAC, using tangential fields with 2.65 Gy per fraction and the total dose prescribed to targeted volume was 42.4 Gy. These patients were observed for acute skin toxicity during the second week and at the end of the treatment. We evaluated dryness, epilation, pigmentation, changes and eritema, dry desquamation (clinically characterized by scaling and pruritus) and moist desquamation (characterized by serious oozing and exposure of the dermis). By using the radiation therapy oncology group’s (RTOG) toxicity criteria, we assessed the skin toxicity over and at the end of the breast irradiation course. Results: 17 patients had eritema gr I and 3 patients haven’t had any skin reaction. There was no patient with great degree of eritema or mois desquamation. In this group of patients, the most frequent skin reaction was eritema gr I.

Conclusion: The hypofractionation radiotherapy of whole breast in the early stage of breast cancer provides a good quality of life because of the short duration of its treatment, its good tolerance of skin and the low grade of skin toxicity related to the treatment.

Keywords – skin toxicity, hypofractionated breast irradiation

1. INTRODUCTION

Radiotherapy is a crucial component in the treatment of breast cancer. Over the past 30 years, several studies showing the survival equivalence of mastectomy and breast conservation (lumpectomy or quadrantectomy with breast radiation) in the treatment of early-stage breast cancer [1–4] have been conducted. We are now able to fully appreciate the impact that the local control has on the overall survival [5]. The recent studies suggest that improvements in local control of greater than 10% at 5 years will likely translate into a 5% improvement in overall survival at 15 years. As we look forward to new techniques for the treatment of breast cancer, we need to be certain that the impact of the local control is not compromised. We know that a whole-breast irradiation (WBI), often followed by a tumor bed boost, for the treatment of early-stage breast cancer is an effective treatment with limited toxicity. As new techniques are adopted, we must avoid compromising these excellent results. In order to preserve our gains in local control, cosmesis and quality of life,
guidelines have been proposed for a patient selection [6] when using some of the new techniques.

This study will focus on the hypofractionation radiotherapy, as one of these new techniques of radiation that are being used when breast-conserving surgery is chosen, and the skin toxicity during a hypofractionated breast irradiation.

1.1. Hypofractionation

Whole-breast irradiation using standard fractionation consisting of 45–50 Gy using daily fraction sizes of 1.8–2.0 Gy over 5 weeks with or without the addition of a tumor bed boost is the most commonly used regimen for early-stage of breast cancer following breast-conserving surgery. Higher daily fraction sizes (hypo fractionation) could offer the advantage of shortening the overall treatment time, making breast preservation possible for a broader population, especially in places where there is poor access to radiation facilities. The concern over the larger fraction sizes is that there may be an increase in the long-term toxicity, potentially leading to an increase in fibrosis, pain and poor cosmetic outcomes.

Recently, there have been three prospective, randomized trials published comparing the standard fractionation with the hypofractionation in the treatment of early-stage of breast cancer [7, 8, 9].

When taking all three of these studies into consideration, it appears that accelerated, hypofractionated regimens for the early-stage of breast cancer should be considered as the standard treatment for many patients with early-stage of breast cancer. These regimens are more economical because they reduce the number of treatment days and have the potential to increase access to breast conservation for patients who may have difficulty getting to a radiation facility or committing to 6 weeks of daily treatment. It is still unclear whether patients with grade 3 tumors will achieve similar local control rates with the accelerated, hypofractionated course, and for the time being, the standard fractionation is probably best outside of a clinical trial setting. It is also unclear how to incorporate boost doses of radiation into the accelerated, hypofractionation regimens, and for now, patients who may benefit from a boost (e.g., young patients or those with close resection margins) should be treated with a standard fractionation, with the addition of a boost.

These skin reactions are categorized as early effects or late effects, depending on the time at which they present.

Early effects are those that occur within 90 days of the initiation of radiation. Those skin reactions occurring during the second to fourth week of therapy include dryness, epilation, pigmentation changes, and erythema [10]. During the third to sixth week of therapy a dry desquamation can develop [11]. The dry desquamation is clinically characterized by scaling and pruritus. Moist desquamation may occur following four to five weeks of therapy [10].

Late effects are those that present more than 90 days after the completion of radiotherapy, and are associated with injury to the dermis. The late effects of atrophy and fibrosis are directly related to a dermal fibroblast response to radiotherapy. Pigmentation changes can also occur as a late reaction. Telangiectasias can develop 6 months to multiple years following the completion of radiotherapy.

Dermal necrosis also can occur months to years following radiotherapy. Necrosis is associated with doses higher than those used to treat the breast [10]. This form of skin injury is related to microvascular changes that result in dermal ischemia [11].

2. MATERIALS AND METHODS

Twenty patients with early stage of breast cancer (Stadium I and II) and conservative surgery (quadrantectomy of breast with ipsilateral axillary dissection) were analysed.

Patients were analysed by stage of disease, tumor size, hormone receptor status, Her 2 status, additional treatment with chemotherapy or hormonotherapy, smoking, presence of seroma before treatment beginning, presence of skin infection or eритema before treatment beginning and breast size.

Patients were treated with 6MV x rays on LINAC, using tangential fields with 2.65 Gy per fraction and the total dose prescribed to target volume was 42.4 Gy.

Patients were followed during the second week and at the end of the treatment and 6 weeks after treatment for acute skin toxicity. We evaluated dryness, epilation, pigmentation changes and eритema, dry desquamation clinically characterized by scaling and pruritus, moist desquamation characterized by serious oozing and exposure of the dermis.

By using the radiation therapy oncology group’s (RTOG) toxicity criteria, we assessed the skin toxicity over and after the treatment.

3. RESULTS

In our group of patients, 10 patients (50%) were with stage I and 10 patients (50%) with stage II of breast cancer. The whole number of patients with tumor less then 2cm in diameter was 10 (50%) and 10 patients have tumor grater then 2 cm in diameter. 13 patients were hormone receptor positive and 7 patients were hormone receptor negative. 15 patients were Her 2 negative and 5 patients have Her 2 positive expression. All patients were non smokers. Two patients (10%) had seroma in the scar region and one patient (5%) had eритema of breast before starting
radiotherapy. None of the patients had large breast size.

Chemotherapy was added in the treatment of 7 patients, and 13 patients had no chemotherapy treatment. 13 patients had hormone therapy.

The evaluation of skin reactions show that 17 patients have eritema gr I and 3 patients haven’t had any skin reaction. There was no patient with great degree of eritema or moist desquamation. The most frequent skin reaction in this group of patients was eritema gr I.

According to the data from some series, an estimated 90% of patients treated with radiotherapy for breast cancer will develop a degree of radiation-induced dermatitis. The severity of the skin reactions during and following breast irradiation is influenced by both treatment-related and patient-related factors.

The treatment-related factors include the fraction size (the dose delivered with each treatment), the total dose delivered, the volume of tissue treated, the type of radiation [12] and the addition of chemotherapy, [13] The patient-related factors include breast size, smoking, axillary lymphocele drainage before treatment, age, and infection of the surgical wound [14].

Generally, the external beam radiotherapy is a well-tolerated treatment. A clinical trial by Fisher et al [15], which prospectively assessed skin toxicity over the course of breast irradiation using Radiation Therapy Oncology Group (RTOG) toxicity criteria, found less than 3% of patients developed grade III toxicity.

In our study we confirmed that hypo fractionation radiotherapy of early breast cancer is well tolerated treatment. The most frequent skin reaction is eritema gr.1, and we don’t have a patient who developed grade III toxicity.

4. CONCLUSION

The hypofractionation radiotherapy of whole breast in the early stage of breast cancer provides a good quality of life because of the short duration of its treatment, its good tolerance of skin and the low grade of skin toxicity related to the treatment.

5. REFERENCES


APPLICATION OF UV AND X-RAY RADIATION FOR REFRESHING OF OLD LATENT FINGERPRINTS IN THE CYANOACRYLATE FUMING TECHNIQUE – PRELIMINARY STUDY

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Abstract – The work described below focuses on enhancement/refreshment of older latent fingerprints onto a nonporous (glass) surface. The aging of the fingerprints is a major problem of the forensic science, because of the compound’s degradation under the influence of a variety of atmospheric parameters, in addition to the time factor. The humidification extent on the latent fingerprints appeared to be one of the critical factors for the deposition of the cyanoacrylate on the finger-marks, since the unsaturated bonds from lipid content from the latent fingerprints serve as active water condensation centers. In the present work we show results from the enhancement of aged latent fingerprint details by activation of the humidity condensation centers with exposition to UV or X-ray radiation prior to their developing into the cyanoacrylate fuming chamber. The aging of the latent fingerprints on glass surfaces with the time for given conditions was established. A term “critical day” was introduced as the day when the detectible details from the latent fingerprints dropped to one half of the initial number of details, recovered from identical fresh latent fingerprint. As the fingerprints were aged to the “critical day”, one third of the samples were exposed to a UV for 15 minutes, another third to 15 minutes of X-ray radiation and the last third was kept unexposed. All the samples were developed into a single cyanoacrylate fuming process. The results showed that in both UV and X-ray cases, a considerable enhancement/refreshment, in approximately 50% of the aged latent fingerprints, has been achieved with only a short term irradiation (15 minutes).

Keywords – latent fingerprints, cyanoacrylate fuming, UV, X-ray.

1. INTRODUCTION

Due to their uniqueness for each human being, fingerprints have been used in judicial systems worldwide as indisputable evidence for the presence of a given person at a given place. The aging of the fingerprints, however, is a major problem of the forensic science, because of the compound’s degradation under the influence of a variety of atmospheric parameters, in addition to the time factor. Once a fingerprint is deposited, its composition is subjected to a change [1-4]. Aging studies and composition variations have been made by various authors [5, 6]. Some have conducted extensive research to establish the dependence of the changes in relative humidity on the effectiveness of the cyanoacrylate fuming technique [5]. The humidification extent on the latent fingerprints appeared to be one of the critical factors for deposition of the cyanoacrylate on the finger-marks, since the unsaturated bonds from lipid content from the latent fingerprints serve as active water condensation centers, and hence induces an increased polyacryl deposit on the printed ridges.

2. MATERIALS AND METHODS

Fingerprints of fingers 2, 3 and 4 of the same person were deposited onto a glass substrate during the 15 days in sequence. The last day all the samples were developed into one single fuming chamber, whereas fingermarks aging from 0-15 days were produced. For the sake of a better contrast, the fingerprints were subjected to a further treatment with magnetic powder. The samples were photographed with a digital camera. Without any previous digital enhancement, the photographs were analyzed in Photoshop, whereas the characteristic points in the carefully chosen elliptical sections were marked and counted as a total number of “readable” characteristic lines/points, as presented on Figure 1. The decrease in the number of lines for each individual fingerprint indicated the aging of the fingerprints over time.
The day in which the number of “readable” characteristic lines dropped to one half from their initial number (sample as-is or sample denoted as day 0) was evaluated as the “critical day”. In the next step of this research, the deposition of the finger-marks onto glass surface and their aging for 6 days (approximately to the critical day) are considered. Some of the samples were let as-is (not intentionally irradiated with any source of radiation) for comparison. Other samples were subjected to 30 minutes of UV radiation from Hg-lamp (UV-A, UV-B and UV-C). The intensity of the UVA radiation at 20 cm distance (approximate distance from the samples) was 0.2 W/m². Also, other samples were exposed to X-ray radiation for 15 minutes whereas the fingerprint from Finger 3 was irradiated with the central part of the beam from a small educational grade X-ray tube operated at 20 μA and 20 kV. All the aforementioned samples were set in the cyanoacrylate (CA) chamber (humidification and fuming) in a single process. The samples were further developed with a magnetic powder, photographed and subjected to a visual analysis in Photoshop, as described previously. The characteristic points/lines were counted with the same methodology and the numbers indicating the enhancements were presented in line-graphs.

3. RESULTS AND DISCUSSION

The curves of the aging of the latent fingerprints, deposited on glass surfaces, are given on Figure 2. The figure shows the aging (decrease of the characteristic lines and artifacts) over time for each individual finger and for the average value of the characteristic points from all three fingerprints is abrupt during the first 5-6 days. It is also evident that the number of the mentioned points drops to one half of its initial value after approximately 6 days. Hence, the day-6 was considered to be the critical day of aging.

Fig. 1 – Counting the number of characteristic points

Fig. 2 - Aging of the fingerprints deposited on a glass surface - number of characteristic lines versus the day of age.
Figure 3 shows the photographs of the developed fingerprints by the CA-fuming and the additional magnetic powder visualization of the prints from fingers 2, 3 and 4, deposited on a glass surface. The left three columns pertain to the images of fingerprints after 6 days upon deposition, whereas the right three columns show the samples which are 15 days old. The top row with the photographs shows the as-is (non-irradiated samples) while the bottom row presents the fingerprinted samples, irradiated with UV before the CA-fuming process.

Figure 4 shows the photographs of the developed fingerprints by the CA-fuming and the magnetic powder visualization of fingers 2, 3 and 4, deposited on glass surfaces, whereas the left two columns pertain to the images of fingerprints after 6 days upon deposition. Moreover, the right two columns show the samples which are 15 days old. The top row with the photographs shows the as-is (non-irradiated samples), while the bottom row presents the fingerprinted samples, irradiated with X-rays before their development in the standard CA-fuming process.

The recovery process of the latent fingerprints is evident from Table 1, which clearly shows the increase in the number of characteristic lines obtained from the aged latent fingerprints due to the previous irradiation with either UV or X-ray sources. From Table 1 it can be seen that about 50% relative enhancement (refreshment) of the mean number of characteristic points for identification in the aged latent fingerprints could be achieved with application of either 30 minutes of UV radiation or 15 minutes of X-ray radiation. In other words, the number of the characteristic points/lines grows for about 50% as the samples are irradiated with either UV or X-rays, prior to their development into the CA-chamber. Figure 5 presents the enhancing effect of the aforementioned irradiation for each separate finger (2, 3 and 4) to each of the examined radiation sources (UV of X-ray).
Fig. 4 - Comparison of fingerprints of Fingers 2, 3 and 4 on glass surfaces, recovered after 6 days (left two columns) and to 15 days (right two columns), as-is (top row) and irradiated with X-ray radiation (bottom row) before the CA fuming.

Table 1. Recovery of the fingerprints - number of characteristic points after and before the irradiation with UV or X-ray.

<table>
<thead>
<tr>
<th></th>
<th>Aged as-is</th>
<th>Irradiated with UV</th>
<th>Irradiated with X-ray</th>
<th>UV-relative enhancement [%]</th>
<th>X-ray-relative enhancement [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Finger 2</td>
<td>14</td>
<td>24</td>
<td>22</td>
<td>41.7</td>
<td>36.4</td>
</tr>
<tr>
<td>Finger 3</td>
<td>3</td>
<td>13</td>
<td>7</td>
<td>76.9</td>
<td>57.1</td>
</tr>
<tr>
<td>Finger 4</td>
<td>6</td>
<td>14</td>
<td>15</td>
<td>57.1</td>
<td>60.0</td>
</tr>
<tr>
<td>Mean of 3 fingers</td>
<td>7.7</td>
<td>17</td>
<td>14.7</td>
<td>54.9</td>
<td>47.7</td>
</tr>
</tbody>
</table>

4. CONCLUSION

The aging of the latent fingerprints on glass surfaces with the time at given conditions was established. At a normal room temperature and atmospheric conditions, the “critical day” (the day when the readable lines/details from the latent fingerprints dropped to one half of the initial number of characteristic lines), was found to be the sixth day. As the fingerprints were aged to the “critical day”, one third of the samples were exposed to a UV for 15 minutes, another third to 15 minutes of X-ray radiation and the last third was kept unexposed. The results showed that in both UV or X-ray cases, a considerable enhancement/refreshment of the aged
latent fingerprints of approximately 50% has been achieved with only a short term irradiation (30 minutes and 15 minutes, correspondingly). Since this is only a preliminary study, the chemical mechanism of the organic changes due to aging, the reversible creation of the condensation centers for an enhanced humidification and the CA-deposition on the old latent fingerprints remain to be studied in the future.

5. REFERENCES


RADIATION DOSIMETERS FOR MEDICAL USE

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Abstract – The several personal radiation dosimeter types for medical use, which look like promising for this kind of application, as pMOS (RADFET) dosimeter, direct ion storage (DIS) dosimeters, thermoluminescent (TL) and optically stimulated luminescent (OSL) dosimeters, are described, and their advantages and disadvantages are analyzed. The p-channel metal-oxide-semiconductor (pMOS) dosimetric transistors allow dose measurements in vivo in real time, and they are especially important for radiotherapy. Direct ion storage (DIS) dosimeters are a hybrid of ion chamber and floating gate MOSFETs (FGMOSFETs), show very high sensitivity. Radiative processes that happen during the exposure of crystal to radiation are classified as prompt luminescence or radioluminescence (RL). In the case of an emission during stimulation, this phenomenon is referred to thermoluminescence or optically stimulated luminescence depending on whether the stimulation source is heat or light. TL and OSL dosimeters are natural or synthetic materials, which the intensity of emitted light is proportional to the irradiation dose.

Keywords – radiation dosimeters, pMOS dosimeters, DIS dosimeters, luminescent dosimeters

1. INTRODUCTION

The chosen dosimeters types belong a new dosimeter generation, looking very promising, as radiation sensitive field – effect - transistors (RADFETs), direct ion storage dosimeters (DISD), and optically stimulated luminescent dosimeters (OSLD), and widely used dosimeter, as thermoluminescent dosimeters (TLDs). They are described, emphasizing their advantages and disadvantages. RADFETs are especially important for radiotherapy radiation since they have extremely small sizes and allow dose measurements in vivo in real time. DIS dosimeter is a hybrid of ion chamber and floating gate MOSFETs (FGMOSFETs), having very high sensitivity. The difference between OSLD and TLD in the stimulation source: heat or light. TLD is mainly used as a personal dosimeter in hospitals, changing the old film dosimeter.

2. DOSIMETERS DESCRIPTION

2.1. pMOS dosimeter (RADFET)

p-channel metal-oxide-semiconductor (pMOS) dosimetric transistors (the second name is Radiation Sensitive Field-Effect-Transistors - RADFETs) are unique radiation dosimeters that have extremely small sizes (the dimensions of sensor elements are ~ 1 mm x 1 mm), and allow dose measurements in vivo in real time, which is specially important for radiotherapy.

The basic concept of pMOS dosimeter is to convert the threshold voltage shift, \( \Delta V_T \), induced by radiation, into absorbed radiation dose, \( D \). This dependence can be expressed in the form: \( \Delta V_T = A \cdot D^n \), where \( \Delta V_T = V_T - V_{T0} \), \( V_T \) is the threshold voltage after irradiation, \( V_{T0} \) before radiation, \( A \) is a constant, and \( n \) is the degree of linearity. \( n \) depends on oxide thickness, electric field and absorbed dose, and, ideally, \( n \) is linear, i.e., \( n = 1 \).

Figure 1 shows the cross section of pMOS transistors with the defects created by radiation. These defects are the oxide charge and interface traps, and their contributions to the threshold voltage shift could be expressed in the form [1]:

\[
\Delta V_T = \pm \frac{q}{C_{ox}} \Delta N_{ox} + \frac{q}{C_{ox}} \Delta N_a, \tag{1}
\]

where \( \Delta N_{ox} \) and \( \Delta N_a \) are the areal densities of oxide charge and interface traps, respectively, \( q \) absolute value of electron charge, and \( C_{ox} \) oxide capacitance per unit area. The signs "+" and "-" are for p-channel and n-channel MOSFETs, respectively. It means that both the oxide charge and the interface states contribute to the increase of absolute threshold voltage shift of pMOS transistors. This is a reason why pMOS transistors, instead of nMOS transistors, are used as radiation dosimeters.
Fig. 1 – Cross section of irradiated pMOS transistor

The increase in sensitivity to radiation is one of the main objectives when designing pMOS dosimetric transistors. This can be achieved by increasing the gate oxide thickness or stacking more transistors. The investigation of pMOS transistors with a thick gate oxide has been intensified because of their large radiation sensitivity. An interesting idea of thick gate oxide fabrication is the ‘sandwich’ structure, consisting of a layer of thermal and a layer of CVD oxide [1].

The pMOS dosimeter advantages, in comparison with other dosimetric systems, include immediate, non-destructive read out of dosimetric information, extremely small size of the sensor element, the ability to permanently store the absorbed dose, wide dose range, very low power consumption, compatibility with microprocessors, and competitive price (especially if cost of the read out system is taken into account). The disadvantages are a need for calibration in different radiation fields (“energy response”), relatively low resolution (starting from about 1 cGy) and nonreusability.

The range of possibilities of pMOS dosimeter practical application is indeed wide: as a personal dosimeter (potentially), in the laboratory, radiation therapy, spacecraft, nuclear equipment and so on. Fig. 2 shows the using of pMOS dosimeter in radiation therapy (\(^{60}\)Co, LINAC, hadron therapy ...). This kind of application is possible since sensors (pMOS dosimetric transistors) are extremely small (~1 mm\(^2\)) and allow the measurement of dose in vivo in real time.

A new concept is implantable wireless pMOS dosimeter [2]. Each detector is composed of a single pMOSFET with 400 nm oxide, produced by Tyndall Institute, a data acquisition chip, a microprocessor and a copper coil, all encapsulated in a glass tube 3.25 mm in diameter and 25 mm in length (Fig. 3). The circuit is powered by a current induced in the coil by an external handheld antenna connected to an RF reader. The dosimeter is passive during irradiation and powered only for measurement of VT. The microprocessor controls both data acquisition and reader–dosimeter communication. A 12-bit analogue-to-digital converter in the data acquisition chip provides 1 mV resolution in the voltage measurements. A computer controls the RF reader and converts the digital signal to a decimal voltage (Fig. 4).

Fig. 2 – An application of pMOS dosimeter in radiotherapy

2.2. Direct ion storage (DIS) dosimeters

Direct ion storage (DIS) dosimeter is a hybrid of ion chamber and floating gate MOSFETs (FGMOSFETs) [3]. Fig. 5 shows the cross sections of FGMOSFET and DIS. Before irradiation, i.e., before dosimeter using the fully charge is collected in the floating gate (FG), and the \(V_T\) is shifted. For this purpose, an uncovered area is formed on the surface of the floating gate of the MOSFET transistor (Fig. 5a). The radiation dose is determined by change which takes place in the charge of the gate.

This principle for information storage is used in a non-volatile FAMOS (Floating gate Avalanche-injected MOS) memory cell, known as EPROMS, EEPROMS and Flash-Memories. In a non-volatile solid state memory cell the information is stored in the form of electronic charge being trapped on the floating gate of a MOSFET transistor. Original memory designs were only used to store digital information (there was either a low amount or a high amount of charge stored to represent one of the two binary digits 0 or 1). In recent years new types of non-volatile memories have been developed and made commercially available to be used for storing analogue information. This means that the amount of charge in each memory cell can be made fully
variable and, therefore, these memory cells can be used to store analogue information directly.

The radiation sensitivity of normal solid state memory cells is inherently too low for use as detectors in radiation protection applications. The main reason for this is that these devices are specifically designed to be as insensitive as possible to ionizing radiation so that they could be used in space, military and nuclear industry applications without damage. In an analogue EEPROM memory cell, the charge on the floating gate can be set to a predetermined level by injecting electrons through the oxide layer. The charge is then permanently stored on the gate because in the normal operating temperature range the electrons have a very low probability of exceeding the energy barriers in the metal–oxide and oxide–silicon interfaces. These types of memory cells are expected to retain the stored charge for hundreds of years.

![Fig. 5 (a) DIS memory cell, (b) DIS memory cell surrounded by a conductive wall](image)

The DIS principle is based on allowing the surface of the floating gate to be in direct contact with the surrounding gas or air. Ionizing radiation incident in the air space produces electron–ion pairs with extremely high mobility and if there is an electric field surrounding the floating gate, these charge carriers can very efficiently be transferred to the gate before recombination occurs. The electric field around the gate is generated by injecting an initial charge on the gate. By surrounding the whole memory structure with a conductive wall, effectively an ion chamber is formed between the wall and the floating gate (Figure 5b). For photon radiation, the interactions take place mainly in the wall and the secondary electrons ionize the air between the wall and the floating gate. For charged particle radiation, if the wall is sufficiently thin, the charged particles are allowed to transfer their energy directly into the air space. Neutron radiation may interact with atomic nuclei in the wall and induce secondary charged particles. The dosimetric characteristics can therefore be adjusted by altering the properties of the wall material, and other gases instead of air could also be used.

The operating voltage of an ionization chamber based on DIS is 25 to 30 V. To obtain electric field strength of the order of 10 kV/m, necessary for linearity up to high dose rates, the distance between electrodes has to be 2 to 3 mm. The extremely high sensitivity of the DIS methods allows, for the use of chamber volumes less than 1 cm³, with a resulting detection limit well below 10 µSv.

Although DIS dosimeter has many similarities to pMOS dosimeter, and many similar characteristics, the main advantages are very high sensitivity and reusability, but disadvantages are difficult using in real-time radiotherapy, and a complicating design.

2.3. TL and OSL dosimeters

Many crystalline materials produce luminescence (emit light) when either heated or stimulated with a visible light spectrum after exposure to ionizing radiation. The general name for these materials is phosphor. The promptly emitted luminescence observed during radiation is called radioluminescence (RL). Since RL is produced continuously when the material is irradiated, information about the irradiation can be obtained in real-time.

During irradiation, the free holes and free electrons are realized in the valence and conductive band, respectively. The free electrons could be captured at the traps (e.g. electronic trap in Fig. 6). Traps are defined as “crystal defects” which are able to capture a charge carrier and release it to its original band. Their energy levels are within the crystal forbidden band (band gap). Some particular defects that can hold both electrons and holes are referred to as recombination centers. A trapped electron will remain so until it is provided with enough stimulation energy to overcome the trap and eventually recombine with a hole at a recombination center. This recombination can result in the emission of light, i.e. luminescence. Several factors compete with the production processes: some hole traps are classified as non-radiative recombination centers (a hole-electron recombination in those traps will not lead to emission of light). Similarly some electron traps are not responsive to thermal or light stimulations (thermally/optically deep traps). Shallow traps on the other hand are unstable at ambient temperature, and may release their trapped electrons even without external stimulation. The electron captured at recombination centre is in exited state firstly, and then is spontaneously released to ground state recombining with the hole and emitting the visible photon.

Radiative processes that happen during the exposure of the crystal to radiation are classified as prompt luminescence or radioluminescence (RL). In case of an emission during stimulation, this phenomenon is
referred to as thermoluminescence (TL) or optically stimulated luminescence (OSL) depending on whether the stimulation source is heat or light. The presence of shallow, deep traps, as well as non-radiative traps are grouped under the term competing processes, because they interfere with the type of luminescence which is of dosimetric interest. In Fig. 6 the processes in TL and OSL dosimeters are shown. The mechanisms (1) and (2) are for TL/OSL, and RL effects, respectively. An electron in conduction band could be relaxed to the valence band directly, emitting photon that is not visible (mechanism (3)). As is could be easily concluded the physical mechanisms for TL and OSL are very similar.

![Mechanisms of TL and OSL dosimeters](image)

**Fig. 6 – Mechanisms of TL and OSL dosimeters**

### 2.3.1. TL dosimeters (TLDs)

TL detectors are natural or synthetic materials, which emit light which intensity is proportional to the dose of irradiation when heated after having been exposed to radiation. For many years the most commonly used TLDs were LiF phosphor doped with Mg and Ti (LiF:Mg,Ti). The commercial names are TLD-100 (produced by Harshaw), and MTS-N produced in Poland. Then, the following dosimeters produced by Harshaw company: LiF:Mg,Cu,P (TLD-100H), CaF₂:Dy/Tm/Mn (TLD-200/TLD-300/TLD-400), Al₂O₃:C (TLD-500). Nowadays LiF:Mg,Cu,P phosphor detector is produced by different laboratories sometimes under different commercial names, for instance, in the Solid Dosimetric Detector and Method Laboratory, Beijing, China (GR-200 or GR-200A), the Institute of Nuclear Physics, Krakow, Poland (MCP-N). In addition, CaF₂:Mn (produced by the Jozef Stefan Institute, Ljubljana, Slovenia), LiB₄O₇ with different dopants (the Institute for Nuclear Science, Vinca, Serbia), a high sensitive LiF:Mg,Cu,Na,Si (KLT-300) phosphor with a low residual signal, good thermal stability and high sensitivity (the Korea Atomic Energy Research Institute, South Korea), and so on. These commercial dosimeters exhibit an emission maximum at 380 – 480 nm, which corresponds to spectral range of common photomultipliers. The luminescence peak is at 180 – 260 °C, making it easy to read. The shape of the TL curve is also very important. For example, LiF-based dosimeters have complex curves (up to ten peaks). A large number of peaks in TL curves complicates the heating procedure. The heating process should include preheating for the depletion of shallow traps and an additional high-temperature annealing for the depletion of deep traps. CaF₂:Mn and Al₂O₃, which are high sensitive to radiation, show the simplest TL curves. Fig. 7 shows typical TL glow curves with (a) one peak, and (b) more peaks.

![Glow Curves](image)

**Fig. 7 – The simple glow curve with (a) one peak, (b) glow curve with four peaks**

### 2.3.2. OSL dosimeters (OSLDs)

Ionizing radiation creates a large amount of electron-hole pairs in the OSL material. A fraction of these carriers are trapped on energy levels (recombination centers) located in the wide band gap of the insulator. Some of this charge will remain trapped for a period of time depending on both the temperature and the activation energy of the traps. Unlike the thermal anneal used for TL materials, an optical stimulation will provide the energy necessary to release the charges. A subsequent radiative recombination, which emits visible photon, may be observed. Quantifying the amount of emitted light makes it possible to evaluate the dose.

The most famous phosphor for OSLDs is Al₂O₃:C (synthetic sapphire) produced by Landauer, USA, which luminescence is characteristic of the oxygen vacancy centers, representing recombination center, introduced by the presence of carbon impurities (in concentrations up to 5000 ppm). The occupancy of an oxygen vacancy center by two electrons gives rise to a neutral F center, whereas occupancy by one electron forms a positively charged F⁺ center. The main luminescence emission occurs around 420 nm and is believed to be caused by:

\[
F^+ + e \rightarrow F^* \rightarrow F^+ + h\nu, \tag{2}
\]

where F* is an excited F center, which decays to the ground state with the emission of a photon at 420 nm. This process is associated with a relaxation time of approximately 35 ms.
In addition, BaFX materials (X = Br, F, Cl) have been used commercially in OSL for many years, especially for x-ray dose imaging (they are using for x-ray digital radiography as storage phosphors). The best OSL material the wavelengths of stimulation light is strictly divided from wavelengths of emission light (Fig. 8).

**Fig. 8 – Principle of OSL dosimeter**

Fig. 9 shows an active OSL sensor [4]. The power consumption is limited to the bias current in the LED during stimulation, that is, 50 mA during a few seconds under a 2 V bias voltage. The whole sensor fits in half a sugar cube volume (4 mm × 4 mm × 7 mm), and it weights no more than 3 g.

**Fig. 9 – One type of OSL active sensor**

The RL/OSL optical fiber dosimeter system shown in Fig. 10, developed at RisØ [5], is real-time system for radiotherapy. It can be divided into three major components: i) a sensor crystal, ii) an optical detection system, and iii) the signal-processing electronics. The sensor Al₂O₃:Ce crystal is connected to the optical system via a plastic fiber. To produce OSL, a green laser beam is focused through a beam splitter and transported via the optical fiber into the Al₂O₃:Ce dosimeter. The OSL signal is then transported back along the same fiber. The dosimeter system is controlled using a standard laptop, and uses only one optical fiber to reduce the size of the probe inserted into a patient for in vivo dosimetry.

A problem is that the signal generated by ionizing radiation in the optical fiber during exposure introduces a noise component in all real-time measurements. Extensive literature attests to the presence of this so-called stem-effect in optical fiber dosimeters using both plastic and silica fibers. The main contributor is Cerenkov radiation, which is generated in a dielectric medium when a charged particle with a velocity greater than that of light in the medium is passing through the medium. In addition, the big lack of system is its emphasized complication.

The RL is collected during irradiation, while the OSL is measured by switching the laser on after the irradiation is completed. This approach allows the investigation of the respective properties of the RL and the OSL signals (see Fig. 11).

Because of their very good characteristics, firstly the high sensitivity, TLDs&OSLDs are irreplaceable in personal, medical and environmental dosimetry.

The advantages of TL&OSL dosimeters relating to others are very high sensitivity, very good linearity to even seven orders of magnitude, re-usability, low energy dependence, low fading (the ability to store dosimetric information for a long time), and reproducibility. The sensitivity threshold is even ~5 – 10 μGy (=μSv for gamma and x photons), which is much bellow 100 μGy required by international standard. Manufacturers of TL&OSL dosimeters claim that they may be re-used at least 2000 times without noticeable change in sensitivity.

**Fig. 20 – The RL/OSL real-time dosimetry system**

The disadvantages are the saturation of response curve after ~ 1 Gy, and impossibility of real time using (for TLDs). TL curves with one isolated peak are desirable; otherwise, if several peaks are present, the dosimeter heating protocol is complicated. Although this is still under debate whether TLD or OSLD is better, it seems that OSLDs have some advantages. Besides real time possibility, the main advantages are ability: 1) of multiple measurements of dose, and 2) to provide imaging information. 1) The measuring of the TL signal completely empties the filled traps, and re-reads are not possible. With optical stimulation, however, one has fine control over the degree to which the traps are emptied and, by varying the intensity and the wavelength of the
stimulation light, multiple measurements are possible—up to 15 independent measurements of the absorbed dose have been demonstrated. It should be emphasized that each re-evaluation of the dose is a completely independent fresh analysis. As a result, dosimeters may be retained and archived for re-analysis at a future date, at the request of the customer or other agency. 2) An additional advantage of the OSL technique is its ability to provide imaging information. One of the most desirable features of film badge technology is the ability to ‘image’ the radiation field. Imaging requires the use of large area detectors in order to allow the spatial variations in the radiation field to be monitored. Spatial imaging is not so feasible with TLDs, which are small in order that they may be heated uniformly.

3. CONCLUSION

The pMOS dosimeter advantages, in comparison with other dosimetric systems, include immediate, non-destructive read out of dosimetric information, extremely small size of the sensor element, the ability to permanently store the absorbed dose, wide dose range, very low power consumption, compatibility with microprocessors, and competitive price (especially if cost of the read out system is taken into account). The disadvantages are a need for calibration in different radiation fields ("energy response"), relatively low resolution (starting from about 1 cGy) and nonreusability.

The DIS dosimeters have many similarities to pMOS dosimeter, and their main advantages are very high sensitivity and reusability, but disadvantages are difficult using in real-time radiotherapy, and a complicating design, as well as relatively high operating voltage (25 - 30 V).

Because of their very good characteristics, firstly the high sensitivity, TLDs&OSLDs are irreplaceable in personal, medical and environmental dosimetry.

The advantages of TL&OSL dosimeters relating to others are very high sensitivity, very good linearity to even seven orders of magnitude, re-usability, low energy dependence, low fading (the ability to store dosimetric information for a long time), and reproducibility.

The disadvantages are the saturation of response curve after ~ 1 Gy, and impossibility of real time using (for TLDs). TL curves with one isolated peak are desirable; otherwise, if several peaks are present, the dosimeter heating protocol is complicated. Although this is still under debate whether TLD or OSLD is better, it seems that OSLDs have some advantages. Besides real time possibility, the main advantages are ability: 1) of multiple measurements of dose, and 2) to provide imaging information.

4. REFERENCES


THE DIGITAL FLAT-PANEL X-RAY DETECTORS

Goran S. Ristić

Abstract – In a digital imaging system, the incident x-ray image must be sampled both in the spatial and intensity dimensions. In the spatial dimension, samples are obtained as averages of the intensity over picture elements or pixels. In the intensity dimension, the signal is digitalized into one of a finite number of levels or bits. Two main types of digital flat-panel detectors are based on the direct conversion, which contains the photoconductor, and on indirect conversion, which contains phosphor. The basics of these detectors are given. Coupling traditional x-ray detection material such as photoconductors and phosphors with a large-area active-matrix readout structure forms the basis of flat panel x-ray imagers. Active matrix technology provides a new, highly efficient, real time method for electronically storing and measuring the product of the x-ray interaction stage whether the product is visible wavelength photons or electrical charges. The direct and indirect detectors, made as the active-matrix flat-panel detectors containing sensing/storage elements, switching elements (diodes or thin film transistors (TFTS)) and image processing module, are described. Strengths and limitations of stimulable phosphors are discussed. The main advantages and disadvantages of mentioned x-ray detectors are also analyzed.

Keywords – X-ray, digital detector, photoconductor, phosphor

1. INTRODUCTION

X-ray film has been most popular in medical imaging. It is cheap, simple to, and readily available in large sheets. Storage phosphors, also known as imaging plates and first commercially developed in Japan in the 1980s, are an alternative to film. In practice, the storage phosphor plate is exposed to x rays and is read out by raster scanning the plate with red laser light. Medical imaging requires that a radiologist be able to see sufficient detail and contrast to make a diagnosis.

A new generation of large-area, flat-panel detectors with integrated, thin-film transistor promises very rapid access to digital images. As digital radiography continues this rapid evolution, it is likely that radiologists will be inundated with information concerning a wide variety of large-area, flat-panel electronic detectors [1-3].

The basis of two digital detector types, direct and indirect detectors, is given. The direct and indirect detectors, made as the active-matrix flat-panel detectors containing sensing/storage elements, switching elements (diodes or thin film transistors (TFTS)) and image processing module, are described. Their main advantages and disadvantages are analyzed. Strengths and limitations of stimulable phosphors are also discussed.

2. DIGITAL X-RAY IMAGING

2.1. Basis of digital x-ray detectors

In a digital imaging system, the incident x-ray image must be sampled both in the spatial and intensity dimensions. In the spatial dimension, samples are obtained as averages of the intensity over picture elements or pixels. These are usually square, and spaced at equal intervals throughout the plane of the image. A fraction of pixel that is sensitive to the incoming signal is the geometrical fill factor. In the intensity dimension, the signal is digitalized into one of a finite number of levels or bits. The pixel size and the bit number must be appropriate chosen for a given imaging task. Each pixel typically contains a switching element and a sensing/storage element.

Coupling traditional x-ray detection materials such as phosphors or photoconductors with a large-area active-matrix readout structure forms the basis of flat-panel x-ray imagers. Active matrix technology provides a new, highly efficient, real time method for electronically storing and measuring the product of the x-ray interaction stage whether the product is visible wavelength photons or electrical charges. Three steps in image creation are: 1) detection, 2) storage, and 3) measurement stages.
Two types of digital flat-panel x-ray detectors (Fig. 1) exist, depending on detection types:
- Direct detection that incorporate a photoconductor to produce electrical charges on detection of an x-ray.
- Indirect detection that incorporate a phosphor to produce visible photons on detection of an x-ray.

**Fig. 1 – Direct and indirect conversion**

For direct conversion a thick (~0.5 to 1 mm) amorphous Se (a-Se) photoconductive layer ($Z = 34$) is usually used. The pixels incorporate a conductive electrode to collect charge and a capacitor element to store it. Interacting x-rays produce charge in the photoconductive which is then shared between the inherent capacitance of the photoconductive layer and the pixel-storage capacitance (Fig. 2).

**Fig. 2 – Cross section of a single pixel with a-Se**

In the band structure for a photoconductor shown in Fig. 3a an optical photon with energy $E_g$ can excite an electron from the valence band to the conduction band leaving behind a hole in the valence band (internal photoelectric effects). The energy of light photons is between 1 and 3 eV, but band gap $E_g \approx 2$ eV for photoconductors ($E_g = 2.2$ eV for a-Se). Else, $E_g \approx 1$ eV for semiconductors (e.g. 1.1 eV for Si).

For the high energetic x-rays having energy thousands of times higher energy than $E_g$, the rules are different. The many materials are using in diagnostic imaging have high atomic number $Z$, and the absorption of diagnostic x-rays is dominant by photoelectric effect. A very energetic electron is liberated, which passes through the material, causes further ionization. Under these circumstances, the amount of energy necessary to create electron-hole pair is not simply $E_g$, but $3E_g$.

**Fig. 3 – The electronic band structures of a) photoconductors & semiconductors, and b) phosphor**

In indirect conversion, a phosphor layer is placed in intimate contact with an active matrix array (Fig. 4). The intensity of the light emitted from a particular location of the phosphor is a measure of the intensity of the x-ray beam incident on the surface of the detector at that point. Each pixel on the active matrix has a photosensitive element that generates an electrical charge whose magnitude is proportional to the light intensity emitted from the phosphor in the region close to the pixel. This charge is stored in the pixel until the active-matrix array is read out.

**Fig. 4 – Cross section of a single pixel with phosphor**

The imaging system is completed with peripheral circuitry that amplifies, digitizes, and synchronizes the readout of the image and a computer that manipulates and distributes the final image to the appropriate soft- or hard-copy devices (Fig. 5).

**Fig. 5 – The layout of pixel groups on active-matrix array**
2.1.1. Photoconductor in direct conversion

The material used as the x-ray photoconductor is not the pure form of a-Se. Pure a-Se is not thermally stable and tends to crystallize over time. However, the crystallization of a-Se can be prevented be alloying a-Se with about 0.5 % As, denotes as a-Si: 0.5 % As.

a-Se advantages are:
- Could be easily and cheaply made in large areas by a low-temperature process,
- Uniform in imaging properties to a very fine scale (an amorphous material is entirely free from granularity),
- There no free carriers at room temperature.

a-Se disadvantages:
- Very high voltage needed to activate a-Si layer (~10 V/μm); under fault conditions, this voltage could damage the active-matrix array,
- Atomic number Z = 34 is rather low and requires very thick layers for high quantum efficiency at diagnostic energies (~100 keV).

Alternatives for photoconductors are: PbI₂, PbO, TiBr, and potentially: CdZnTe, CdTe, CdSe and HgI₂.

2.1.2. Phosphor in indirect conversion

A phosphor is common name for a material that glows after exposure to radiation. Many current x-ray imaging detectors employ a phosphor in the initial stage to absorb x rays and produce light. Phosphor work by exciting electrons from the valence band to the conduction band where they are free to move a small distance within the phosphor (Fig. 3b). Some of these electrons will decay back to the valence band through a local state created by small amounts of impurities called activators, emitting the light (Fig. 3b). Else, the electrons will be firstly trapped at excited state of activator site (so-called recombination centre), and then be recombined with hole reached from valence band emitting the light.

Thus, phosphor can be relatively efficient converters of the large incident energy of the x-ray into light photons. Because light photons each carry only small (~2-3 eV) energy, many light photons are created from the absorption of a single x-ray. This quantum amplification is the conversion gain of the phosphor.

For instance, CsI:Tl during irradiation produces the light photons with wavelength of 600 nm corresponding to energy of 2 eV. If it is irradiated by 60 keV, 30 000 light photons should be expected. However, energy necessary for a light photon production is not 2 eV, but 18 eV (~3Eg; Eg = 6.2 eV for CsI), and 3300 light photons are produced per a x ray. Else, the band gaps for phosphors are from 5 to 10 eV.

One of the main issues with the phosphor is the balance between spatial resolution and x-ray detection. The thicker phosphor the more x-ray is absorbed, the spatial resolution is worst, since the emitted light can spread further from the point of production before existing the screen. The conflict could be solved using a needle like structure (structured phosphor, similar to optic fibers), as CsI phosphor (83 % of the emitted light will undergo internal reflection).

The besides main advantage of very high resolution comparing to other phosphor types the main disadvantages of CsI are: hydroscopic nature, toxicity, and lack of mechanical robustness.

2.2. Active-matrix flat-panel detectors

The first active-matrix flat-panel display was demonstrated by Brody in 1973, and was fabricated using CdSe as the semiconductor. The majority of today’s flat-panel arrays are fabricated from hydrogenated amorphous silicon (a-Si:H). The role of hydrogen is to occupy many of the dangling bonds present in raw a-Si. One significant advantage of a-Si:H for medical imaging flat-panel active-matrix devices is that the layers of a-Si:H can be deposited over extremely large areas (exceeding 1x1 m²).

The main parts of active-matrix array pixels are:
1) Sensing/storage elements, and
2) Thin-film switching elements.

1) Sensing and storage elements are photodiodes for the indirect approach and storage elements are capacitors for the direct approach.

2) Thin-film switching elements are:
- two-terminal devices, as diodes, metal-insulator-metal devices (MIMs), metal-semiconductor-isolator devices (MSIs), …
- three-terminal devices, as thin field transistors (TFT).

2.2.1. Sensing/storage elements for indirect detection

The photodiodes as sensing elements in indirect pixel are designed to detect visible lights. The photodiodes also acts as a capacitor to store the photogenerated charge. The absorption coefficient of a-Si:H (~ 104 to 106 cm⁻¹) for visible lights is an order of magnitude higher than that for crystalline silicon (~ 103 to 105 cm⁻¹) even though the effective optical band gap of a-Si:H (1.7 eV) is larger than that in crystalline silicon (1.1 eV). A 0.5 µm-thick layer of undoped (i.e., interstitial) a-Si:H is sufficient to absorb the most lights entering the layer, but often layer thickness of the order of 1 to 2 µm may be used to reduce pixel capacitance (Fig. 6).

Fig. 7 shows the absolute quantum efficiency as a function of photon wavelength in the visible range for an ~1.5 µm thick n-i-p photodiode at -5 V reverse bias. Also in figure are the emission spectra for typical x-ray phosphors. It can bee seen that the absorption of a-Si:H is well matched to the emission from these phosphors.
2.2.2. Sensing/storage elements for direct detection

In the direct detection approach, the photoconductor is the sensing element which converts incoming x-ray into charge. A simple charge storage capacitor is the storage element that requires only dielectric and metal layers. The storage capacitor is electrically connected to the overlying photoconductor at pixel electrode.

A bias voltage must be applied across the thickness of the photoconductor to facilitate the separation and collection of x-ray induced charge. To maintain an internal field of 10 V/μm within a 500-μm-thick layer of a-Se, a bias of 5000 V must be applied. Three methods incorporate a means of draining away the potential on the pixel if it exceeds a predetermined safe design value (Fig. 8):

- including an extra component parallel with the storage capacitor (e.g., a Zener diode),
- modifying the TFT by incorporating of a second gate connected to the pixel electrode to ensure that its channel current will increase to drain away the excess charge when the pixel potential approaches damaging levels,
- reversing the a-Se structure to permit a negative bias on the top electrode to ensure that the ordinary TFT will start to conduct when the pixel potential reaches the threshold voltage.

2.2.3. Switching elements: diodes

The other main component of the pixels on an active-matrix array is the switching element, and there are a number of different possibilities for its design. One is using diodes for both the sensing and switching elements, and the diodes are usually of the same type (i.e., either all Schottky or all n-i-p diodes). Fabricating both diodes at the same time reduces the number of mask levels and consequently improves the device yield. Advantage is easy fabrication, and disadvantages are: create an image lag (offset image) and high capacitance.

2.2.4. Switching elements: thin film transistor (TFT)

The switching devices with the best properties for the majority of medical imaging applications is the a-Se:H thin film transistor (TFT; see Fig. 9). Silicon nitride (a-Si3N4:H) is the usual gate dielectric. The positive gate voltage (+10 to +15 V) is used to switch device on and a negative voltage (−5 to −10 V) to switch it off. The on to off ratio for the current through this device is extremely high (> 1010), so that they have excellent switching properties. A radiation part is absorbed by TFT, and although it should be radiation hard, the TFT threshold voltages are usually changed.

2.2.5. Image processing and manipulation

Because of limited tolerances achievable in controlling the thickness and quality of the different layers on large-area flat panel x-ray imagers, the sensitivity of the pixels to the radiation and their offset signals (from dark current integration and switching transients) vary from pixel to pixel. This includes variations resulting from the thickness and quality of the photoconductor or phosphor layers coupled to the arrays. Tolerance issues in the fabrication of the peripheral amplification and controlling circuitry also add variations in the pixel sensitivity and offset. The most powerful method for
Variations in pixel and electronic offsets are corrected using dark-field images acquired with no x-ray exposure. Pixel sensitivity and electronic gain variations are corrected using flood-field images taken with a constant intensity of x-ray exposure across the full area of the detector. Measured drifts of pixel sensitivity are extremely small, even over extended periods of time, so acquisition of flood field data is not needed as frequently as offset correction. A carry over or lag or ghosting may be observed after large exposures, and it may sometimes be necessary corrected. This effect is much more emphasized in direct than in indirect detectors.

2.3. The photostimulable (storage) phosphor

One of the most popular and most successful detectors for digital radiography to date have been photo stimulable phosphors, also known as storage phosphors. These phosphors are commonly in the barium fluorohalide family, typically BaFBr:Eu+2, where the atomic energy levels of the europium activator determine the characteristics of light emission. X-ray absorption mechanisms are identical to those of conventional phosphors. They differ in that the useful optical signal is not derived from the light that is emitted in prompt response to the incident radiation, but rather from subsequent emission when electrons and holes are released from traps in the material (Fig. 10).

The photostimulable phosphor is an excellent detector for digital radiography since, when placed in a cassette, it can be used with conventional x-ray machines. Large-area plates are conveniently produced, and images can be acquired quickly.

The plates are reusable, have linear response over a wide range of x-ray intensities, and are erased simply by exposure to a uniform stimulating light source to release any residual traps. One limitation of this type of detector is that because the traps are located throughout the depth of the phosphor material, the laser beam providing the stimulating light must penetrate into the phosphor. Scattering of the light within the phosphor causes release of traps over a greater area of the image than the size of the incident laser beam spot. This results in loss of spatial resolution, which is emphasized if the plate is made thicker to increase x-ray absorption.

The main advantage of this system, comparing to the flat-panel detectors, is reusability. The storage phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some phosphor may be used a lot of times without an information losing. In the flat panel detector the some pixels are damaged by radiation and ghost image will appear after certain time. The main disadvantage is
very high price of image reader, and impossibility of real-time imaging (fluoroscopy).

2.4. Digital system for fluoroscopy

Fig. 12 shows an x-ray image intensifier (XRII) that is currently only one digital system for fluoroscopy that represent real-time visualization. The resulting real-time images are usually displayed using a video system (conventional or CCD) optically coupled to the x-ray image intensifier. The XRII absorbs the incident x-ray image, amplifies and outputs it as an optical image which is then distributed by lenses to the video camera. X-rays are converted to light in the large input phosphor screen typically of 12.5 cm to 40 cm in diameter. The fluorescence illuminates a photocathode evaporated directly on the phosphor and liberates electrons. The purpose of the photocathode is to convert light photons to electrons efficiently. The electrons are accelerated through a large potential difference (typically 25 kV) and electrostatically focused by the electrodes onto a small (2.5 cm diameter) output phosphor.

The main disadvantages of digital systems based on the use of XRIIs are loss of image contrast due to x-ray and light scatter within the tube, the geometric distortion of the image due to the curved input phosphor, and an influence of earth’s magnetic field. The main advantages are: could be easily and cheaply made in large areas by a low-temperature process, uniform in imaging properties to a very fine scale (an amorphous material is entirely free from granularity), and no free carriers at room temperature. It disadvantages are very high voltage needed to activate a-Si layer (~10 V/μm), which could damage the active-matrix array, low atomic number (Z = 34), requiring very thick layers for high quantum efficiency. Promising alternatives for a-Se could be: PbI₂, PbO, TlBr, and potentially: CdZnTe, CdTe, CdSe, and HgI₂.

In the case of phosphor, there is the balance between spatial resolution and x-ray detection. The thicker phosphor the more x-ray is absorbed, the spatial resolution is worst, since the emitted light can spread further from the point of production before exiting the screen. The conflict could be solved using a needle like structure (similar to optic fibers), as CsI phosphor. The besides main advantage of very high resolution comparing to other phosphor types the main disadvantages of CsI are: hydroscopic nature, toxicity, and lack of mechanical robustness.

The main advantage of photostimulable phosphor system, comparing to the flat-panel detectors, is reusability. The storage phosphor may be used a lot of times without an information losing. In the flat panel detector the some pixels are damaged by radiation and ghost image will appear after certain time. The main disadvantage is very high price of image reader, and impossibility of real-time imaging (fluoroscopy).

3. CONCLUSION

There are two main types of digital flat-panel x-ray detectors that are based on:
- Direct detection that incorporate a photoconductor to produce electrical charges on detection of an x-ray.
- Indirect detection that incorporate a phosphor to produce visible photons on detection of an x-ray.

For direct conversion a thick (~0.5 to 1 mm) amorphous Se (a-Se) is usually used. The pixels incorporate a conductive electrode to collect charge and a capacitor element to store it.

In indirect conversion, a phosphor layer is placed in intimate contact with an active matrix array.

a-Se advantages are: could be easily and cheaply made in large areas by a low-temperature process, uniform in imaging properties to a very fine scale (an amorphous material is entirely free from granularity), and no free carriers at room temperature. It disadvantages are very high voltage needed to activate a-Si layer (~10 V/μm), which could damage the active-matrix array, low atomic number (Z = 34), requiring very thick layers for high quantum efficiency. Promising alternatives for a-Se could be: PbI₂, PbO, TlBr, and potentially: CdZnTe, CdTe, CdSe, and HgI₂.

4. REFERENCES

HOW USEFUL ARE INDEPENDENT MU CALCULATIONS IN A HIGHLY AUTOMATIZED RADIOTHERAPY CENTRE

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Abstract – A number of national and international medical physics organizations recommend implementing independent MU verification (MUV) as a part of a comprehensive QA program. AAPM Report 114 is coherent with these recommendations. The authors of this report explain why should MUV be performed in every clinic. They divide the clinics in three groups, according to their sophistication in performing planning and treatment. The third group is classified as a fully automated clinic. In such a clinic all parameters needed for MU calculations are transferred from the primary treatment planning system. In that case, we cannot expect to obtain different result than that obtained from the primary TPS. We discuss how useful independent routine MU calculation is in such clinic. Our conclusion is that for routine treatment plans, made according to protocols, there is no need to perform MUV. For non-routine and unusual treatment plans, checks of the plans could be made and that decision will be made by a medical physicist. When introducing a new treatment modality and a new treatment planning scheme, the physicists should make measurements in order to verify that the TPS calculated MU are correct for delivery of the prescribed dose. In vivo dosimetry should be performed in all cases for the first treatment of the patient.

Keywords – monitor units, treatment protocols, radiotherapy.

1. INTRODUCTION

Independent monitor units (or time) calculations are part of the overall QA process in radiation therapy. Over the last few decades, the method of independent MU calculations was the primary method for catching treatment planning errors. In January 2011, the American Association of Physicists in Medicine has issued the report of its task group 114 titled: Verification of monitor units calculations for non-IMRT clinical radiotherapy: Report of AAPM Task Group 114 [1]. In the introductory part of the report they establish that ‘Modern treatment planning systems (TPSs), ... are complex. This complexity present challenges to traditional manual verification methods since much of the information required to perform the verification needs to come directly from the TPS.’ Despite the fact that the monitor units verification (MUV) can detect error in the planning process, ‘the MUV is not a check of the accuracy of the entire calculated dose distribution’.

In the report, they also divide radiation therapy centers as follows: Three groups of centers regarding connectivity among parts of equipment: 1) Manual, 2) Partial connectivity, and 3) Full integration. They further conclude that ‘The majority of centers likely fall into category (2) with many centers approaching category (3)’.

In this paper we deal exclusively with the statements and recommendations from the report [1]. AAPM is influential professional organization and their standings have significant influence all over the world. We will try to explain that category 3 centers, i.e. centers with full integration, should take other approaches to verify MU than the recommended calculational approach suggested by AAPM. As a primer we will take our center which belongs to group 3. Single quotes in this paper represent citations from the AAPM report.

2. MATERIALS AND METHODS

There is huge variability amongst radiotherapy centers regarding their equipment, methods of work, preference of therapy execution and so on. But the availability of radiotherapy equipment and other supporting equipment is the main factor of determining sophistication of MU calculation and treatment delivery.

In order to present the level of sophistication of treatments in our institution we will depict in several words the process of patient treatment in our center.
The patient is scanned on a CT scanner and the CT slices are electronically via network transferred to the treatment planning system (Eclipse). There, from the slices, a 3D model of the patient is built. On this 3D model we are making conformal plan putting appropriate fields in BEV mode. After that, we calculate the MU and if unsatisfied with the isodose distribution we make adequate changes to the plan. This is iterative process and we proceed until we are satisfied with the isodose distribution and the fulfillment of the treatment prescription requirements. According to these working procedures and connectivity among equipment, our department belongs to the third group of radiotherapy centers defined in the AAPM report.

We will present several arguments described in the AAPM report. These arguments have shaking validity regarding MUV for a modern fully integrated radiation therapy center. They are self-navigating towards our conclusions of MU checking. However, the AAPM report sticks to the conclusion that the primary method always should be independent calculation of MU.

In the report it is stressed that modern TPS are very sophisticated and that their commissioning is very tedious, time consuming and complex and that, in fact, it is impossible to check the algorithms for every possible treatment situation. Because of that ‘subtle changes in process could lead to the use of the untested modules within the TPS or usage in a manner that was not originally tested’. Certainly, such a situation could be potentially dangerous for patients. But yet, taking into account that wrong calculated MU are a gross error they conclude that ‘an effective MUV is one of the several tools in the QA process designed to catch gross errors in the treatment dose delivery for an individual patient’.

Maybe one of the crucial suggestion in this report is that ‘Hospital and departmental administrators must provide adequate physics staffing to allow for the timely completion of the patient plan QA checks, including MUV.’ This suggestion is not directly connected with MUV but it is of essential importance for providing conditions for proper QA check of every patient plan. Connected with this remark, they ascertain that ‘It is recognized that most centers do not have the resources to perform a comprehensive set of tests of TPS on their own’. Of course, this can be argument to insist on MUV for every patient but still it is worse decision than to invest every effort to make a comprehensive set of tests of the TPS.

One requirement for a MUV system to be efficient in detecting planning errors is to satisfy the establishment that ‘The verification program typically uses a different beam and/or patient model than the primary TPS’. The authors of reference [2] have made comparisons of MU calculations between a TPS and commercially available independent monitor unit verification software. They determined the differences in MU calculations between two systems but there was no conclusion about found planning errors.

We came to a point where the report recommendation forces us to make MUV for every treatment plan we calculate. Usually these are routine, protocol based plans and they are several hundred a year. In the report the authors conclude that going towards group 3 (centers with full integration) huge changes are introduced in modern radiotherapy practice and ‘This changes have brought into the question the value of performing the MUV’. Thus, we run against a paradox.

3. RESULTS

There are solutions to this paradox. For the nonstandard treatment plans and for new conceived planning scheme one solution could be to make simulations which convergence will show that the calculations of tretment planning system are correct or uncorrect. The other type of MU verifications could be through experimental measurements.

Fig. 1 – Algorithm for choosing whether to make MU verification or not

In-vivo measurement should be used besides and additionally to experimental measurements to check whether TPS correctly computes in that new treatment situation.

In Fig. 1 we present the algorithm which can be followed in order to decide whether to perform independent MU calculation or not. As it is shown on this flow chart, when we work standard protocol there is no need for MUV. If we make some nonstandard plan (out of the existing protocols) the physicist should decide whether MUV should be made or not; but usually, in most cases, the check ought to be made. It should be stressed that these nonstandard plans are pretty rare in our clinic, in any case less than 10% of all patients treated with radiotherapy.

Additionally, the report states that ‘Measurement is most useful when the difference between the primary and the verification calculations is outside the action level’. But this statement is valid for the third case in our flow chart and it is natural to perform measurements along with calculations in the process of protocol establishment. So, in this process, we will polish our MUV algorithm to know its limitations and restrictions.
If the hospital and departmental administrators stick to the recommendations of AAPM generally and to this report particularly, with the adequate staff the radiotherapy center can fulfill above mentioned three methods. That will be sufficient to avoid errors which may appear in newly devised treatment methods. It should be emphasized that this suggested approach to reduce independent MUV will be time consuming (for the new planning protocols) and physics staff demanding.

4. DISCUSSION

It seems that one argument for recommendation to perform MUV is the evidence that most centers do not have the resources to perform a comprehensive set of tests of TPS on their own. But in that case these centers won’t have enough resources to perform proper MUV too. So, regular MUV for every patient is not substitution for a comprehensive commissioning of TPS and other equipment.

With the process of MUV mainly random errors can be caught, as is declared in the report: ‘Systematic errors are best reduced by a thorough review of the process and a careful commissioning procedure’. These random errors, as some of them are counted in the report, cannot be caught by MUV because they will be transferred from the TPS to the independent program for MUV.

Regarding a fully integrated radiotherapy center it is questionable to define the term independent MU calculations. In the circumstances of such a center, all initial input parameters for MU calculations are taken from the dataset already present and unchangeable in the patient database. Then, there is no probability that MUV, although based on different beam parameters and data, will reveal result significantly different than that from the TPS.

5. CONCLUSION

The method of MUV should be used highly selectively, in a department in which all radiotherapy and supporting equipment are highly connected. In order not to waste precious time, the plans that are made strictly following a protocol need not to be checked for MU errors. When a new kind of treatment plan is introduced, the physics staff should make comprehensive measurements which will approve that the MU calculations of TPS are correct. For precautions several next plans can be checked for MU errors by independent MU calculation. After that, there will be no need to make further MUV for the next plans. Regular QA of all equipment will prevent and discover unwanted events (like corruption of data in TPS) and will be the correct way of obviate any equipment malfunction. In all cases in vivo dosimetry should be performed at least for the first treatment.

6. REFERENCES


ADVANTAGES OF THE TECHNIQUE WITH SEGMENTED FIELDS FOR TANGENTIAL BREAST IRRADIATION

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Abstract - In the case of breast cancer, the prominent role of radiation therapy is an established fact. Depending on the stage of the disease, the breast is most often irradiated with two tangential fields and a direct supraclavicular field. Planning target volume is defined through the recommendations in ICRU Reports 50 and 62. The basic ‘dogma’ of radiotherapy requires the dose in the target volume to be homogeneous. The favorable situation would be if the dose width was between 95% and 107%; this, however, is often not possible to be fulfilled. A technique for enhancement of homogeneity of isodose distribution would be using one or more additional fields, which will increase the dose in the volume where it is too low. These fields are called segmented fields (a technique also known as ‘field in field’) because they occupy only part of the primary fields. In this study we will show the influence of this technique on the dose homogeneity improvement in the PTV region. The mean dose in the target volume was increased from 49.51 Gy to 50.79 Gy in favor of the plans with segmented fields; and the dose homogeneity (measured in standard deviations) was also improved - 1.69 vs. 1.30. The increase in the target volume, encompassed by 95% isodose, was chosen as a parameter to characterize overall planning improvement. Thus, in our case, the improvement of dose coverage was from 93.19% to 97.06%.

Keywords – tangential fields, segmented fields, radiotherapy, dose homogeneity

1. INTRODUCTION

The radiotherapy techniques in the treatment of breast cancer vary in different institutions, but, in general, the issue of radiation dose delivery to the chest wall after total mastectomy or to the breast following breast conservation surgery remains complex.

Radiotherapy treatment fields are usually tangential to encompass the breast or thoracic wall, and, in some cases, matched to a supraclavicular field. A target volume usually has an irregular shape and because of that it is generally difficult to achieve homogeneous isodose distribution. In order to overwhelm this inhomogeneity, wedge filters are usually used to decrease the dose in the dorsal part of the breast while increasing the dose in the ventral part. In such a way it is possible sometimes to homogenize the dose in acceptable boundaries.

CT-based three-dimensional conformal treatment planning allows the planner and physician to evaluate the dosimetry across the entire breast. According to ICRU-50 guidelines, an optimal plan is one in which the entire planning target volume (PTV) is between 95% and 107% levels relative to 100% prescription point [1]. The main difficulty with tangential field breast irradiation is in the achievement of homogeneous dose distribution inside the target volume. Low dose volumes within the target volume of the breast and the thoracic wall result in reduced tumor control probability, and the magnitude of this effect depends on the amount of macroscopic and microscopic residual disease; whereas high dose volumes could lead to increased late normal tissue morbidity. Retrospective data show that the reduction of dose from 50 Gy to 45 Gy may lead to reduction in local control from 95% to 85%. In those cases where inhomogeneity is a serious concern, especially for large breasts, planning with field-in-field technique may reduce inhomogeneity dose regions inside the target volume.

In order to improve the isodose distribution, the techniques of segmented fields have been introduced. The techniques of segmented fields (or field in field technique) were enabled due to advent of multileaf collimators and computerized treatment planning systems. A number of authors describe this technique as it is implemented in their institutions [2, 3]. The present work aims to compare the dose delivery parameters of the isocentric field-in-field technique plan vis-à-vis other simple plans with parallel opposed tangential fields in the treatment of breast cancer.
2. MATERIALS AND METHODS

For the purpose of implementing standard 3D conformal radiation therapy the patients were simulated on a CT simulator, immobilized on an inclined breast board. The patients were scanned in spiral mode with a distance of 0.5 cm between the slices. Using these slices in the TPS Eclipse we made a 3D reconstruction of the scanned part of the patient’s body. The isocenter was determined during this phase of simulation. Radiation oncologists contour planning target volume (PTV) and for the sake of the treatment report they contour planning target volume for evaluation (PTV_EVAL). This evaluation volume goes 5 mm below the skin. Also, glandular tissues, infraclavicular and supraclavicular nodes (ICLN, SCLN), are contoured as tissues which should receive certain dose (at least 40 Gy). As organs of risk, the ipsilateral lung is contoured and also the heart, in case of left breast irradiation. 

The target volumes were defined and the dose prescribed according to the International Commission on Radiation Units and Measurement (ICRU) Reports 50 and 62 recommendations. Accordingly, the target volume should be surrounded by the 95% isodose line. The planning target volume (PTV) definition for the breast and chest wall was done according to the breast cancer atlas for radiation therapy planning consensus definitions of the Radiation Therapy Oncology Group (RTOG).

The fundamental technique of tangential fields consists of two fields (6MV photon beams pseudo-opposite fields): one of the fields enters the breast or thoracic wall medially with the edge on the sternum and the opposite edge goes 1.5-2 cm in air flushing above the skin surface of the breast or thoracic wall. The other, or lateral, field with one edge enters axillary on the line where medial field exits and its opposite edge flash over the most prominent part of breast for 1.5-2 cm. According to this, the gantry angles are chosen in such a way that the contralateral breast should be out of open fields and as less as possible volume of lung and heart as organs at risk to be encompassed in irradiated volume. Multileaf collimator (MLC) is used where the lung and the heart could be shielded. We accept that maximum dose in PTV should be around 107% and the isodose distribution should be symmetrical in all axial planes regarding the medial and lateral breast side. This localization usually has an irregular volume and shape, and because of that it is generally difficult to achieve homogeneous isodose distribution. To overwhelm this inhomogeneity, wedge filters are usually used to decrease the dose in the dorsal part of the breast while increasing the dose in the ventral part. In such a way it is sometimes possible to homogenize the dose in acceptable boundaries.

But, very often, despite the use of the wedge filters, a lot of target volume is subdosed i.e. there are regions in the target volume PTV_EVAL with doses much less than 95% and also very unsymmetrical. In this case, we used the technique of segmented fields to try...
to improve the isodose distribution. Segmented fields treated in this work are always under gantry angles of either medial tangential field or lateral tangential field.

The techniques of segmented fields (or field in field technique) were enabled due to the advent of multileaf collimators and computerized treatment planning systems. The technique of segmented fields used in our clinic consists of the following procedures. First, we optimize the isodose distribution as much as possible on the classical plan. After that, we display only 95% and 100% isodose lines in a 3D view mode. In this mode we can see which parts of the target volume PTV_EVAL are underdosed. Next, we copy one of the tangential fields, usually the medial field, and paste it as an open field. The weight factor for this field will be 0 in order not to disturb isodose distribution and also we put MLC on the field. In BEV, with the help of MLC, we shield the volume which receives more than 100%. We increase the weight factors for the initial fields by \((n+1)/n\) ratio (\(n\) is the initial number of fields) and the weight factor for the newly created segmented field will be assigned casually about 0.2 or 0.3. We calculate MU and inspect the isodose distribution, and we adjust the weight factor for the segmented field in such a way as to get a bit less percentage dose in the open field than in the shielded field region. In such a case we make point dose as a dose which should be a parameter that should ensure comparability of the treatment plans without and with segmented fields. At our department, we strive to achieve the number of MU for segmented field to be greater than 10. If it is not possible to fulfill that criterion we don’t accept the segmented field. We can see in figure 1. the implementation of the technique of segmented fields in three steps.

For the purposes of this work, we recorded minimum dose, maximum dose, mean dose and standard deviation of the dose for the PTV_EVAL volume, both for the technique of classical tangential fields and for technique with segmented fields; and maximum dose, mean dose and standard deviation for the lung tissue in both cases too. Also, we recorded the percentage doses which encompass 95% of the PTV_EVAL volume in both cases.

### 3. RESULTS

Twenty eight pseudo-consecutive patients treated in 2011 were investigated and several dose parameters were compared for plans with classical tangential fields and for plans with segmented field(s) added. Several patients were excluded from this study, mainly because their thoracic walls were too thin for implementing the technique of segmented fields.

Because the plans with segmented fields are built upon the classical technique we use, the samples (without and with segmented fields) are dependent. Paired t Test for the mean of PTV_EVAL for classical plan and segmented plan shows superior values for segmented plans. The mean values of chosen parameters differ at 0.05 level of significance in favor of plans with segmented fields and particularly the mean of EVAL(mean) is significantly different from the mean of segmented fields cases EVAL_s(mean) \((p<0.05)\). This statistical conclusion is also valid for the minimum value of dose in PTV_EVAL and for the standard deviation of dose in this volume. The results of the calculated doses are shown in table 1. Boxplots for the means of target dose for both cases are shown in figure 2.

<table>
<thead>
<tr>
<th></th>
<th>Classical</th>
<th>Segmented</th>
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<tbody>
<tr>
<td>EVAL(mean)</td>
<td>49.51</td>
<td>50.79</td>
</tr>
<tr>
<td>EVAL(min)</td>
<td>42.18</td>
<td>43.53</td>
</tr>
<tr>
<td>EVAL(STD)</td>
<td>1.69</td>
<td>1.30</td>
</tr>
<tr>
<td>Lung(max)</td>
<td>49.75</td>
<td>50.35</td>
</tr>
<tr>
<td>Lung(STD)</td>
<td>15.22</td>
<td>15.28</td>
</tr>
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</table>

For lung tissue, the mean dose does not have appropriate meaning, so we use maximum dose in lung and the standard deviation of the dose in lung to represent the change in dose distributions. The mean of maximum dose in lung for the case of segmented fields is significantly different from the mean of maximum dose in lung for the traditional case of tangential fields \((p=0.008)\). But this little increase of maximum dose does not significantly influence the standard deviation of dose change in lung. It reveals that with \(p=0.609\) there is not enough evidence to conclude that the means of standard deviations differ at the 0.05 level of significance.

In a way, it is self-evident and expected that segmented field(s) will contribute to better dose distribution in the target volume and this statistical analysis shows that situation. In order to obtain some numerical parameter to characterize the evident improvement in dose distribution, we compare the isodose levels that encompass 95% of the target volume in both cases. 95% of target volume (PTV_EVAL) receiving 95% of prescribed dose has more medical meaning too; therefore, this approach has a direct link to the clinical expectations of the treatment. The statistical analysis shows that the mean of the isodose level for the segmented field cases is significantly different from the mean of the isodose level for traditional technique with tangential fields. In all patients but four, we managed to encompass more than 95% of the target volume with the 95% isodose curve, Table 2. shows the values of these isodose levels. The level of significance was taken to be 0.05 as in all other calculations.
4. DISCUSSION

In order to achieve a greater dose of homogeneity, the use of the segmented fields in a number of treatment planning occasions was proved to be of great value.

<table>
<thead>
<tr>
<th>Mean</th>
<th>93.19</th>
<th>97.06</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard Deviation</td>
<td>3.11</td>
<td>1.91</td>
</tr>
</tbody>
</table>

Table 2. Values of the isodose levels that encompass 95% of the target volume in cases without segmented fields ($D(V95)$eval) and in cases with segmented fields ($Ds(V95)$eval).

Fig. 2 Boxplots showing the distribution of mean dose values for PTV EVAL for plans with tangential fields alone and for plans with segmented field(s) included.

This is also true for treating breast cancer. The most commonly used technique for breast cancer radiation therapy treatment is the use of tangential fields. For achieving a better dose homogeneity for classical 3D-CRT, the treatment planner usually uses wedge filters for beam modulation. Nevertheless, on occasion, the planner should use the technique of segmented fields to improve dose homogeneity in some parts of the target volume. Sometimes this improvement could be quite significant. Statistical analysis shows that a mean target dose can be increased for more than 1.2 Gy, minimum target dose can be increased by around 1.3 Gy too, while maximum lung dose will be increased negligibly.

It is expected that the segmented field(s) will contribute to a better dose distribution in the target volume and in that way improve the dose homogeneity. But we need some numerical parameter to characterize the evident improvement in dose distribution. The authors in reference [5] devise a so called PTV Dose Improvement (PDI) index in order to evaluate the effect of their technique. Their method is based on the relative difference between target volumes encompassed between 97% and 103% isodose curves for open fields and method chosen to make treatment plans (IMRT, 3D-CRT, FiF). But that is artificial comparison because open fields are not used for treatment.

5. CONCLUSION

Segmented field technique, or field in field technique, can be considered as IMRT technique obtained with forward planning. It requires from the planner to conceive the favorite isodose distribution in order to implement segmented field later in the course of the planning process. This technique can be successfully used for improving isodose distribution while irradiating breast with tangential fields. As a parameter which can describe the effect of segmented fields technique could be used percentage of target volume encompassed by 95% isodose curve. With segmented fields this 95% - 95% criterion can be fulfilled in the most cases. Certainly, an implementation of the segmented field technique is more time consuming. Actually it adds additional planning time to the planning of a plan with tangential fields but it is rewarded with much better isodose distribution and dose homogeneity in the target volume.

6. REFERENCES


